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(54) Title: METHODS OF OBTAINING OPHTHALMIC LENSES PROVIDING THE EYE WITH REDUCED ABERRAT

(57) **Abstract:** The present invention discloses methods of obtaining ophthalmic lens capable of reducing the aberrations of t comprising the steps of characterizing at least one corneal surface as a mathematical model, calculating the resulting aberr of said corneal surface(s) by employing said mathematical model, selecting the optical power of the intraocular lens. Fro information, an ophthalmic lens is modeled so a wavefront arriving from an optical system comprising said lens and corneal obtains reduced aberrations in the eye. Also disclosed are ophthalmic lenses as obtained by the methods which are capable re aberrations of the eye.

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Methods of obtaining ophthalmic lenses providing the eye with reduced aberrations

Field of invention

The present invention relates to methods of designing ophthalmic lenses that provide the eye with reduced aberrations, as well as lenses capable of providing such visual improvements.

Background of the invention

It is presently discussed that the visual quality of eyes having an implanted intraocular lens (IOL) is comparable with normal eyes in a population of the same age. Consequently, a 70 year old cataract patient can only expect to obtain the visual quality of a non-cataracteous person of the same age after surgical implantation of an intraocular lens, although such lenses objectively have been regarded as optically superior to the natural crystalline lens. This result is likely to be explained by the fact that present IOLs are not adapted to compensate for defects of the optical system of the human eye, namely optical aberrations. Age-related defects of the eye have recently been investigated and it is found that contrast sensitivity significantly declines in subjects older than 50 years. These results seem to comply with the above-mentioned discussion, since the contrast sensitivity measurements indicate that individuals having undergone cataract surgery with lens implantation lens will not obtain a better contrast sensitivity than non-cataracteous persons of an average age of about 60 to 70 years.

Even if intraocular lenses aimed at substituting the defective cataract lens and other ophthalmic lenses, such as conventional contact lenses, have been developed with excellent optical quality as isolated elements, it is obvious that they fail to correct for a number of aberration phenomena of the eye including age-related aberration defects.

US Patent No. 5,777,719 (Williams et al.) discloses a method and an apparatus for accurately measuring higher order aberrations of the eye as an optical system with wavefront analysis. By using a Hartmann-Shack wavefront sensor, it is possible to measure higher order aberrations of the eye and use such data to find compensation for

these aberrations and thereby obtain sufficient information for the design of an optical lens, which can provide a highly improved optical performance. The Hartmann-Shack sensor provides means for analyzing light reflected from a point on the retina of the eye of a subject. The wavefront in the plane of the pupil is recreated in the plane of the lenslet array of the Hartmann-Shack sensor. Each lenslet in the array is used to form an aerial image of the retinal point source on a CCD camera located at the focal plane of the array The wave aberration of the eye, in the form resulting from a point source produced on the retina by a laser beam, displaces each spot by an amount proportional to the local slope of the wavefront at each of the lenslets. The output from the CCD camera is sent to a computer, which then performs calculations to fit slope data to the first derivatives of 66 Zernike polynomials. From these calculations, coefficients for weighting the Zernike polynomials are obtained. The sum of the weighted Zernike polynomials represents a reconstructed wavefront distorted by the aberrations of the eye as an optical system. The individual Zernike polynomial terms will then represent different modes of aberration.

US Patent No. 5,050,981 (Roffman) discloses another method for designing a lens by calculating modulation transfer functions from tracing a large number of rays through the lens-eye system and evaluating the distribution density of the rays in the image position. This is repeatedly performed by varying at least one lens surface until a lens is found which results in a sharp focus and a maximum modulation transfer function.

US Patent No. 6,224,211 (Gordon) describes a method of improving the visual acuity of the human eye by successively fitting aspheric lenses to the cornea and thereby finding a lens that can reduce spherical aberration of the whole individual eye.

The methods referred to above for designing are suitable for the design of contact lenses or other correction lenses for the phakic eye which can be perfected to compensate for the aberration of the whole eye system. However, to provide improved intraocular lenses aimed to replace the natural crystalline lens, it would be necessary to consider the aberrations of the individual parts of the eye.

US Patent No. 6,050,687 (Bille et al) refers to a method wherein the refractive properties of the eye are measured and wherein consideration is taken to the contribution of the individual surfaces of the eye to the total wavefront aberrations. The method

described herein particularly aims at analyzing the topography of the posterior corneal surface in order to improve refractive correction techniques.

There has recently been a focus on studying the aberrations of the eye, including number of studies of the development of these aberrations as a function of age. In two particular studies, the development of the components of the eye were examined separately, leading to the conclusion that the optical aberrations of the individual components of younger eyes cancel each other out, see Optical Letters, 1998, Vol. 23(21), pp.1713-1715 and IOVS, 2000, Vol.41(4), 545. The article of S. Patel et al in Refractive & Corneal Surgery, 1993, Vol. 9, pages 173-181 discloses the asphericity of posterior corneal surfaces. It is suggested that the corneal data can be used together with other ocular parameters to predict the power and the asphericity of an intraocular lens with the purpose of maximizing the optical performances of the future pseudophakic eye Furthermore, it was also recently observed by Antonio Guirao and Pablo Artal in IOVS, 1999, Vol. 40(4), S535 that the shape of the cornea changes with age and becomes more spherical. These studies indicate that the cornea in the subjects provides a positive spherical aberration, which increases slightly with the age. On the other hand, the rotationally symmetric aberration of the anterior corneal surface does not seem to be different between younger and older eye according to results found by T Oshika et al in Investigative Ophthalmology and Visual Science, 1999, Vol. 40, pp. 1351-1355.In Visic Research, 1998, 38(2), pp. 209-229, A Glasser et al. investigated the spherical aberratioı of natural crystalline lenses from eyes obtained from an eye bank after the cornea has been removed. According to the laser scanner optical method used herein it was found that the spherical aberration from an older lens (66 years) shows positive spherical aberration, whereas a 10-year-old lens shows negative spherical aberration. In addition, Vision Research, 2001, 41, pp.235-243 (G Smith et al) discloses that the natural crystalline lens appears to have negative spherical aberration when in the relaxed state. Smith et al suggest that because older eyes have a larger aberration, it is likely that the spherical aberration of the crystalline lens becomes less negative with age.

In Ophthal. Physiol. Opt., 1991, Vol. 11, pp. 137-143 (DA Atchison) it is discussed how to reduce spherical aberrations in intraocular lenses by aspherizing the lens surface. The methods outlined by Atchison are based on geometric transferring

calculations, which do not consider diffraction effects and any variations in refractive index along the ray path in inhomogeneous elements. These calculations will lead to errors close to the diffraction limit. Also in WO 98/31299 (Technomed) a ray tracing method is outlined according to which the refraction of the cornea is attempted to be considered for the design of an intraocular lens.In view of the foregoing, it is apparent that there is a need for ophthalmic lenses that are better adapted or compensated to the aberrations of the individual surfaces of the eye and are capable of better correcting aberrations other than defocus and astigmatism, as provided by conventional ophthalmic lenses.

Description of the invention

It is an object of the invention to provide for methods that result in obtaining an ophthalmic lens, which provides the eye with reduced aberrations.

It is another object of the invention to provide methods of obtaining an intraocula lens capable of reducing the aberration of the eye after its implantation into the eye.

It is a further object to provide for methods of obtaining an intraocular lens capable of compensating the aberrations resulting from optical irregularities in the corneal surfaces.

It is a still further object of the present invention to provide an intraocular lens which is capable of restoring a wavefront deviating from sphericity into a substantially more spherical wavefront.

It is a also an object of the present invention to provide an intraocular lens which is capable of correcting for mean optical irregularities and imperfections found in a particular group of people and thereby provide a lens with improved optical performance for an individual belonging to the same group.

The present invention generally relates to an ophthalmic lens and to methods of obtaining said ophthalmic lens that is capable of reducing the aberrations of the eye. By aberrations in this context is meant wavefront aberrations. This is based on the understanding that a converging wavefront must be perfectly spherical to form a point image, i.e. if a perfect image shall be formed on the retina of the eye, the wavefront

having passed the optical surfaces of the eye, such as the cornea and a natural or artifici lens, must be perfectly spherical. An aberrated image will be formed if the wavefront deviates from being spherical. . In this context the term nonspherical surface will refer t rotationally symmetric, asymmetric and/or irregular surfaces, i.e. all surfaces differing from a sphere. The wavefront aberrations can be expressed in mathematical terms in accordance with different approximate models as is explained in textbook references, such as M.R. Freeman, Optics, Tenth Edition, 1990.

In a first embodiment, the present invention is directed to a method of designing an intraocular lens capable of reducing aberrations of the eye after its implantation. The method comprises a first step of characterizing at least one corneal surface as a mathematical model and by employing the mathematical model calculating the resulting aberrations of the corneal surface. An expression of the corneal aberrations is thereby obtained, i.e. the wavefront aberrations of a spherical wavefront having passed such a corneal surface. Dependent on the selected mathematical model different routes to calculate the corneal aberrations can be taken. Preferably, the corneal surface is characterized as a mathematical model in terms of a conoid of rotation or in terms of polynomials or a combination thereof. More preferably, the corneal surface is characterized in terms of a linear combination of polynomials. The second step of the method is to select the power of the intraocular lens, which is done according to conventional methods for the specific need of optical correction of the eye, for example the method described in US Patent No. 5,968,095 From the information of steps one anc two an intraocular lens is modeled, such that a wavefront from an optical system comprising said lens and corneal model obtains reduced aberrations. The optical system considered when modeling the lens typically includes the cornea and said lens, but in th specific case it can also include other optical elements including the lenses of spectacles or an artificial correction lens, such as a contact lens, a corneal inlay implant or an implantable correction lens depending on the individual situation.

Modeling the lens involves selection of one or several lens parameters in a syste which contributes to determine the lens shape of a given, pre-selected refractive power. This typically involves the selection of the anterior radius and surface shape, posterior radius and surface shape, the lens thickness and the refractive index of the lens. In

practical terms, the lens modeling can be performed with data based on a conventional spherical lens, such as the CeeOn® lenses from Pharmacia Corp., as exemplified with the CeeOn® Edge (Model 911). In such a case, it is preferred to deviate as little as possible from an already clinically approved model. For this reason, it may be preferred to maintain pre-determined values of the central radii of the lens, its thickness and refractive index, while selecting a different shape of the anterior and/or posterior surface, thus providing one or both of these surfaces to have an nonspherical shape. According to an alternative of the inventive method, the spherical anterior surface of the conventional starting lens is modeled by selecting a suitable aspheric component. Preferably the lens has at least one surface described as a nonsphere or other conoid of rotation. Designing nonspherical surfaces of lenses is a well-known technique and can be performed according to different principles and the description of such surfaces is explained in more detail in our parallel Swedish Patent Application 0000611-4 to which is given reference.

The inventive method can be further developed by comparing wavefront aberrations of an optical system comprising the lens and the model of the average cornea with the wavefront aberrations of the average cornea and evaluating if a sufficient reduction in wavefront aberrations is obtained. Suitable variable parameters are found among the above-mentioned physical parameters of the lens, which can be altered so as to find a lens model, which deviates sufficiently from being a spherical lens to compensate for the corneal aberrations.

The characterization of at least one corneal surface as a mathematical model and thereby establishing a corneal model expressing the corneal wavefront aberrations is preferably performed by direct corneal surface measurements according to well-known topographical measurement methods which serve to express the surface irregularities of the cornea in a quantifiable model that can be used with the inventive method. Corneal measurements for this purpose can be performed by the ORBSCAN® videokeratograph, as available from Orbtech, , or by corneal topography methods, such as EyeSys® from Premier Laser Systems. Preferably, at least the front corneal surface is measured and more preferably both front and rear corneal surfaces are measured and characterized and expressed together in resulting wavefront aberration terms, such as a linear combination of polynomials which represent the total corneal wavefront aberrations. According to one

important aspect of the present invention, characterization of corneas is conducted on a selected population with the purpose of expressing an average of corneal wavefront aberrations and designing a lens from such averaged aberrations. Average corneal wavefront aberration terms of the population can then be calculated, for example as an average linear combination of polynomials and used in the lens design method. This aspect includes selecting different relevant populations, for example in age groups, to generate suitable average corneal surfaces. Advantageously, lenses can thereby be provided which are adapted to an average cornea of a population relevant for an individual elected to undergo cataract surgery or refractive correction surgery including implantation of an IOL or corneal inlays. The patient will thereby obtain a lens that give the eye substantially less aberrations when compared to a conventional spherical lens.

Preferably, the mentioned corneal measurements also include the measurement c the corneal refractive power. The power of the cornea and the axial eye length are typically considered for the selection of the lens power in the inventive design method.

Also preferably, the wavefront aberrations herein are expressed as a linear combination of polynomials and the optical system comprising the corneal model and modeled intraocular lens provides for a wavefront having obtained a substantial reductic in aberrations, as expressed by one or more such polynomial terms. In the art of optics, several types of polynomials are available to skilled persons for describing aberrations. Suitably, the polynomials are Seidel or Zernike polynomials. According to the present invention Zernike polynomials preferably are employed.

The technique of employing Zernike terms to describe wavefront aberrations originating from optical surfaces deviating from being perfectly spherical is a state of th art technique and can be employed for example with a Hartmann-Shack sensor as outlined in J. Opt. Soc. Am., 1994, Vol. 11(7), pp. 1949-57. It is also well established among optical practitioners that the different Zernike terms signify different aberration phenomena including defocus, astigmatism, coma and spherical aberration up to higher aberrations. In an embodiment of the present method, the corneal surface measurement results in that a corneal surface is expressed as a linear combination of the first 15 Zernike polynomials. By means of a raytracing method, the Zernike description can be transformed to a resulting wavefront (as described in Equation (1)), wherein Z_i is the i-tl

Zernike term and a_i is the weighting coefficient for this term. Zernike polynomials are a set of complete orthogonal polynomials defined on a unit circle. Below, Table 1 shows the first 15 Zernike terms and the aberrations each term signifies.

$$z(\rho,\theta) = \sum_{i=1}^{15} a_i Z_i \qquad (1)$$

In equation (1), ρ and θ represent the normalized radius and the azimuth angle, respectively.

Table 1

i	$Z_i(\rho,\theta)$	
1	1	Piston
2	$2\rho\cos\theta$	Tilt x
3	$2\rho\sin\theta$	Tilt y
4	$\sqrt{3}(2\rho^2-1)$	Defocus
5	$\sqrt{6}(\rho^2\sin 2\theta)$	Astigmatism 1st order (45
6	$\sqrt{6}(\rho^2\cos 2\theta)$	Astigmatism 1st order (0
7	$\sqrt{8}(3\rho^3-2\rho)\sin\theta$	Coma y
8	$\sqrt{8}(3\rho^3-2\rho)\cos\theta$	Coma x
9	$\sqrt{8}(\rho^3\sin 3\theta)$	Trifoil 30°
10	$\sqrt{8}(\rho^3\cos 3\theta)$	Trifoil 0°
11	$\sqrt{5}(6\rho^4-6\rho^2+1)$	Spherical aberration
12	$\sqrt{10}(4\rho^4-3\rho^2)\cos 2\theta$	Astigmatism 2nd order (0
13	$\sqrt{10}(4\rho^4-3\rho^2)\sin 2\theta$	Astigmatism 2nd order (45
14	$\sqrt{10}(\rho^4\cos 4\theta)$	Tetrafoil 0°
15	$\sqrt{10}(\rho^4\sin 4\theta)$	Tetrafoil 22.5°

Conventional optical correction with intraocular lenses will only comply with th fourth term of an optical system comprising the eye with an implanted lens. Glasses, contact lenses and some special intra ocular lenses provided with correction for astigmatism can further comply with terms five and six and substantially reducing Zernike polynomials referring to astigmatism.

The inventive method further includes to calculate the wavefront aberrations resulting from an optical system comprising said modeled intraocular lens and cornea ar expressing it in a linear combination of polynomials and to determine if the intraocular lens has provided sufficient reduction in wavefront aberrations. If the reduction in wavefront aberrations is found to be insufficient, the lens will be re-modeled until one o: several of the polynomial terms are sufficiently reduced. Remodeling the lens means tha at least one lens design parameter is changed. These include the anterior surface shape and central radius, the posterior surface shape and central radius, the thickness of the len and its refractive index. Typically, such remodeling includes changing the shape of a ler surface so it deviates from being a spherical. There are several tools available in lens design that are useful to employ with the design method, such as OSLO version 5 see Program Reference, Chapter 4, Sinclair Optics 1996. The format of the Zernike polynomials associated with this application are listed in Table 1.

According to a preferred aspect of the first embodiment, the inventive method comprises expressing at least one corneal surface as a linear combination of Zernike polynomials and thereby determining the resulting corneal wavefront Zernike coefficients, i.e. the coefficient of each of the individual Zernike polynomials that is selected for consideration. The lens is then modeled so that an optical system comprisin; of said model lens and cornea provides a wavefront having a sufficient reduction of selected Zernike coefficients. The method can optionally be refined with the further step of calculating the Zernike coefficients of the Zernike polynomials representing a wavefront resulting from an optical system comprising the modeled intraocular lens and cornea and determining if the lens has provided a sufficient reduction of the wavefront Zernike coefficients of the optical system of cornea and lens; and optionally re-modeling said lens until a sufficient reduction in said coefficients is obtained. Preferably, in this

aspect the method considers Zernike polynomials up to the 4th order and aims to sufficiently reduce Zernike coefficients referring to spherical aberration and/or astigmatism terms. It is particularly preferable to sufficiently reduce the 11th Zernike coefficient of a wavefront front from an optical system comprising cornea and said modeled intraocular lens, so as to obtain an eye sufficiently free from spherical aberration. Alternatively, the design method can also include reducing higher order aberrations and thereby aiming to reduce Zernike coefficients of higher order aberration terms than the 4th order.

When designing lenses based on corneal characteristics from a selected population, preferably the corneal surfaces of each individual are expressed in Zernike polynomials describing the surface topography and therefrom the Zernike coefficients of the wavefront aberration are determined. From these results average Zernike wavefront aberration coefficients are calculated and employed in the design method, aiming at a sufficient reduction of selected such coefficients. In an alternative method according to the invention, average values of the Zernike polynomials describing the surface topography are instead calculated and employed in the design method. It is to be understood that the resulting lenses arriving from a design method based on average values from a large population have the purpose of substantially improving visual quality for all users. A lens having a total elimination of a wavefront aberration term based on an average value may consequently be less desirable and leave certain individuals with an inferior vision than with a conventional lens. For this reason, it can be suitable to reduce the selected Zernike coefficients only to certain degree of the average value.

According to another approach of the inventive design method, corneal characteristics of a selected population and the resulting linear combination of polynomials, e.g. Zernike polynomials, expressing each individual corneal aberration can be compared in terms of coefficient values. From this result, a suitable value of the coefficients is selected and employed in the inventive design method for a suitable lens. In a selected population having aberrations of the same sign such a coefficient value can typically be the lowest value within the selected population and the lens designed from this value would thereby provide improved visual quality for all individuals in the group compared to a conventional lens.One embodiment of the method comprises selecting a

representative group of patients and collecting corneal topographic data for each subjec in the group. The method comprises further transferring said data to terms representing the corneal surface shape of each subject for a preset aperture size representing the pup: diameter. Thereafter a mean value of at least one corneal surface shape term of said gro is calculated, so as to obtain at least one mean corneal surface shape term. Alternativel or complementary a mean value of at least one to the cornea corresponding corneal wavefront aberration term can be calculated. The corneal wavefront aberration terms ar obtained by transforming corresponding corneal surface shape terms using a raytrace procedure. From said at least one mean corneal surface shape term or from said at least one mean corneal wavefront aberration term an ophthalmic lens capable of reducing sai at least one mean wavefront aberration term of the optical system comprising cornea an lens is designed.

In one preferred embodiment of the invention the method further comprises designing an average corneal model for the group of people from the calculated at least one mean corneal surface shape term or from the at least one mean corneal wavefront aberration term. It also comprises checking that the designed ophthalmic lens compensates correctly for the at least one mean aberration term. This is done by measuring these specific aberration terms of a wavefront having traveled through the model average cornea and the lens. The lens is redesigned if said at least one aberration term has not been sufficiently reduced in the measured wavefront.

Preferably one or more surface descriptive (asphericity describing) constants ar calculated for the lens to be designed from the mean corneal surface shape term or from the mean corneal wavefront aberration terms for a predetermined radius. The spherical radius is determined by the refractive power of the lens.

The corneal surfaces are preferably characterized as mathematical models and tl resulting aberrations of the corneal surfaces are calculated by employing the mathematical models and raytracing techniques. An expression of the corneal wavefror aberrations is thereby obtained, i.e. the wavefront aberrations of a wavefront having passed such a corneal surface. Dependent on the selected mathematical model different routes to calculate the corneal wavefront aberrations can be taken. Preferably, the corne surfaces are characterized as mathematical models in terms of a conoid of rotation or in

terms of polynomials or a combination thereof. More preferably, the corneal surfaces are characterized in terms of linear combinations of polynomials.

In one embodiment of the invention, the at least one nonspheric surface of the len is designed such that the lens, in the context of the eye, provides to a passing wavefront a least one wavefront aberration term having substantially the same value but with opposite sign to a mean value of the same aberration term obtained from corneal measurements of a selected group of people, to which said patient is categorized. Hereby a wavefront arriving from the cornea of the patient's eye obtains a reduction in said at least one aberration term provided by the cornea after passing said lens. The used expression 'in the context of the eye' can mean both in the real eye and in a model of an eye.
In a specific embodiment of the invention, the wavefront obtains reduced aberration terms expressed in rotationally symmetric Zernike terms up to the fourth order. For this purpose, the surface of the ophthalmic lens is designed to reduce a positive spherical aberration term of a passing wavefront. The consequence of this is that if the cornea is a perfect lens and thus not will give rise to any wavefront aberration terms the ophthalmic lens will provide the optical system comprising the cornea and the ophthalmic lens with a negative wavefront spherical aberration term. In this text positive spherical aberration is defined such that a spherical surface with positive power produces positive spherical aberration. Preferably the lens is adapted to compensate for spherical aberration and more preferably it is adapted to compensate for at least one term of a Zernike polynomial representing the aberration of a wavefront, preferably at least the 11^{th} Zernike term, see Table 1.

The selected groups of people could for example be a group of people belonging to a specific age interval, a group of people who will undergo a cataract surgical operation or a group of people who have undergone corneal surgery including but not limited to LASIK (laser in situ keratomileusis), RK (radial keratotomy) or PRK (photorefractive keratotomy). The group could also be a group of people who have a specific ocular disease or people who have a specific ocular optical defect.

The lens is also suitably provided with an optical power. This is done according to conventional methods for the specific need of optical correction of the eye. Preferably the refractive power of the lens is less than or equal to 30 diopters. An optical system

considered when modeling the lens to compensate for aberrations typically includes the average cornea and said lens, but in the specific case it can also include other optical elements including the lenses of spectacles, or an artificial correction lens, such as a contact lens, a corneal inlay or an implantable correction lens depending on the individu situation.

In an especially preferred embodiment the ophthalmic lens is designed for peopl who will undergo a cataract surgery. In this case it is has been shown that the average cornea from such a population is represented by a prolate surface following the formula:

$$z = \frac{(1/R)r^2}{1+\sqrt{1-(\frac{1}{R})^2(cc+1)r^2}} + adr^4 + aer^6$$

wherein,

(i) the conical constant cc has a value ranging between −1 and 0

(ii) R is the central lens radius and

(iii) ad and ae are aspheric polynomial coefficients additional to the conical constant.

In these studies the conic constant of the prolate surface ranges between about −0.05 for an aperture size (pupillary diameter) of 4 mm to about −0.18 for an aperture siz of 7 mm. Accordingly an ophthalmic lens suitable to improve visual quality by reducing at least spherical aberration for a cataract patient based on an average corneal value will have a prolate surface following the formula above. Since cornea generally produces a positive spherical aberration to a wavefront in the eye, an ophthalmic lens for implantation into the eye will have negative spherical aberration terms while following the mentioned prolate curve. As will discussed in more detail in the exemplifying part of the specification, it has been found that an intraocular lens that can correct for 100% of a mean spherical aberration has a conical constant (cc) with a value of less than 0 (representing a modified conoid surface), with an exact value dependent on the design pupillary diameter and the selected refractive power. For example, a 6 mm diameter aperture will provide a 22 diopter lens with conical constant value of about − 1.03.

In this embodiment, the ophthalmic lens is designed to balance the spherical aberration c a cornea that has a Zernike polynomial coefficient representing spherical aberration of th

wavefront aberration with a value in the interval from 0.000156 mm to 0.001948 mm for a 3mm aperture radius, 0.000036 mm to 0.000448 mm for a 2 mm aperture radius, 0.0001039 mm to 0.0009359 mm for a 2,5 mm aperture radius and 0.000194 mm to 0.00365 mm for a 3,5 mm aperture radius using polynomials expressed in OSLO format. These values were calculated for a model cornea having a single surface with a refractive index of the cornea of 1.3375. It is possible to use optically equivalent model formats of the cornea without departing from the scope of the invention. For example multiple surface corneas or corneas with different refractive indices could be used. The lower values in the intervals are here equal to the measured average value for that specific aperture radius minus one standard deviation. The higher values are equal to the measured average value for each specific aperture radius plus three standard deviations. The used average values and standard deviations are shown in tables 8,9,10 and 11. The reason for selecting only minus one SD (= Standard Deviation) while selecting plus three SD is that in this embodiment it is convenient to only compensate for positive corneal spherical aberration and more than minus one SD added to the average value would give a negative corneal spherical aberration.

According to one embodiment of the invention the method further comprises the steps of measuring the at least one wavefront aberration term of one specific patient's cornea and determining if the selected group corresponding to this patient is representative for this specific patient. If this is the case the selected lens is implanted and if this is not the case a lens from another group is implanted or an individual lens for this patient is designed using this patients corneal description as a design cornea. These method steps are preferred since then patients with extreme aberration values of their cornea can be given special treatments.

According to another embodiment, the present invention is directed to the selection of an intraocular lens of refractive power, suitable for the desired optical correction that the patient needs, from a plurality of lenses having the same power but different aberrations. The selection method is similarly conducted to what has been described with the design method and involves the characterizing of at least one corneal surface with a mathematical model by means of which the aberrations of the corneal surface is calculated. The optical system of the selected lens and the corneal model is

then evaluated so as to consider if sufficient reduction in aberrations is accomplished by calculating the aberrations of a wavefront arriving from such a system. If an insufficient correction is found a new lens is selected, having the same power, but different aberrations. The mathematical models employed herein are similar to those described above and the same characterization methods of the corneal surfaces can be employed.

Preferably, the aberrations determined in the selection are expressed as linear combinations of Zernike polynomials and the Zernike coefficients of the resulting optical system comprising the model cornea and the selected lens are calculated. From the coefficient values of the system, it can be determined if the intraocular lens has sufficiently balanced the corneal aberration terms, as described by the Zernike coefficients of the optical system. If no sufficient reduction of the desired individual coefficients are found these steps can be iteratively repeated by selecting a new lens of the same power but with different aberrations, until a lens capable of sufficiently reducing the aberrations of the optical system is found. Preferably at least 15 Zernike polynomials up to the 4^{th} order are determined. If it is regarded as sufficient to correct for spherical aberration, only the spherical aberration terms of the Zernike polynomials for the optical system of cornea and intraocular lens are corrected. It is to be understood that the intraocular lens shall be selected so a selection of these terms become sufficiently small for the optical system comprising lens and cornea. In accordance with the present invention, the 11^{th} Zernike coefficient, a_{11}, can be substantially eliminated or brought sufficiently close to zero. This is a prerequisite to obtain an intraocular lens that sufficiently reduces the spherical aberration of the eye. The inventive method can be employed to correct for other types of aberrations than spherical aberration by considering other Zernike coefficients in an identical manner, for example those signifying astigmatism, coma and higher order aberrations. Also higher order aberrations can be corrected dependent on the number of Zernike polynomials elected to be a part of the modeling, in which case a lens can be selected capable of correcting for higher order aberrations than the 4^{th} order.

According to one important aspect, the selection method involves selecting lenses from a kit of lenses having lenses with a range of power and a plurality of lenses within each power having different aberrations. In one example the lenses within each power

have anterior surfaces with different aspherical components. If a first lens does not exhibit sufficient reduction in aberration, as expressed in suitable Zernike coefficients, then a new lens of the same power, but with a different surface (aspheric component) is selected. The selection method can if necessary be iteratively repeated until the best lens is found or the studied aberration terms are reduced below a significant borderline value. In practice, the Zernike terms obtained from the corneal examination will be directly obtained by the ophthalmic surgeon and by means of an algorithm they will be compared to known Zernike terms of the lenses in the kit. From this comparison the most suitable lens in the kit can be found and implanted. Alternatively, the method can be conducted before cataract surgery and data from the corneal estimation is sent to a lens manufacture for production of an individually tailored lens.

The present invention further pertains to an intraocular lens having at least one nonspherical surface capable of transferring a wavefront having passed through the cornea of the eye into a substantially spherical wavefront with its center at the retina of the eye. Preferably, the wavefront is substantially spherical with respect to aberration terms expressed in rotationally symmetric Zernike terms up to the fourth order.

In accordance with an especially preferred embodiment, the invention relates to an intraocular lens, which has at least one surface, when expressed as a linear combination of Zernike polynomial terms using the normalized format, that has a negative 11^{th} term of the fourth order with a Zernike coefficient a_{11} that that can balance a positive such term of the cornea „ to obtain sufficient reduction of the spherical aberration of the eye after implantation. In one aspect of this embodiment, the Zernike coefficient a_{11} of the lens is determined so as to compensate for an average value resulting from a sufficient number of estimations of the Zernike coefficient a_{11} in several corneas. In another aspect, the Zernike coefficient a_{11} is determined to compensate for the individual corneal coefficient of one patient. The lens can accordingly be tailored for an individual with high precision.

The invention further relates to another method of providing a patient with an intraocular lens, which at least partly compensates for the aberrations of the eye. This method comprises removing the natural lens from the eye. Surgically removing the impaired lens can be performed by using a conventional phacoemulsification method.

The method further comprises measuring the aberrations of the aphakic eye, not comprising a lens, by using a wavefront sensor. Suitable methods for wavefront measurements are found in J.Opt.Soc.Am., 1994, Vol. 11(7), pp. 1949-57 by Liang et. a Furthermore, the method comprises selecting from a kit of lenses a lens that at least part compensates for the measured aberrations and implanting said lens into the eye. The kit of lenses comprises lenses of different power and different aberrations and finding the most suitable lens can be performed in a manner as earlier discussed. Alternatively, an individually designed lens for the patient can be designed based on the wavefront analys of the aphakic eye for subsequent implantation. This method is advantageous, since no topographical measurements of the cornea are and the whole cornea, including the front and back surfaces, is automatically considered.

The lenses according to the present invention can be manufactured with conventional methods. In one embodiment they are made from soft, resilient material, such as silicones or hydrogels. Examples of such materials suitable for foldable intraocular lenses are found in US patent No. 5,444,106 or in US Patent No. 5,236,970. Manufacturing of nonspherical silicone lenses or other foldable lenses can be performed according to US Patent No. 6,007,747. Alternatively, the lenses according to the present invention can be made of a more rigid material, such as poly(methyl)methacrylate. The skilled person can readily identify alternative materials and manufacturing methods, which will be suitable to employ to produce the inventive aberration reducing lenses.

Detailed description of the invention

Fig.1 shows a comparison of the a_{11} ("Z11") Zernike coefficient values for 10 subjects if implanted with CeeOn® 911 lenses and the inventive averaged ("Z11") lens.

Fig. 2 shows modeled visual acuities of the test subjects with CeeOn® 911 lenses and the inventive averaged ("Z11") lenses.

Fig. 3 and Fig. 4 show modulation transfer function comparisons between CeeOn® 911 lenses and the inventive averaged ("Z11") lenses

Fig. 5 shows visual acuity plotted as a function of the astigmatism of the lenses according to the model lenses according to the invention.

Fig. 6 shows the best corrected visual acuity with the inventive lenses.

Fig. 7 and 8 show modulation transfer functions of an individual with an individually designed lens.

Fig. 9 shows the best corrected visual acuity with individually designed lenses according to the invention.

Fig. 10 shows the age distribution of 71 patients used in a study described below in the example part.

Fig. 11 shows a height map given by an Orbscan® true height data file.

Fig. 12 shows average corneal wavefront aberration coefficients.

Fig. 13 shows a scatter plot of the spherical aberration of 71 subjects for a 6 mm diameter aperture.

Fig. 14 shows a scatter plot of the spherical aberration of 71 subjects for a 4 mm diameter aperture.

Fig. 15 shows a scatter plot of the spherical aberration of 71 subjects for a 5 mm diameter aperture.

Fig. 16 shows a scatter plot of the spherical aberration of 71 subjects for a 7 mm diameter aperture.

Example 1

A sample set of 10 corneal surfaces from individuals were described using Zernike polynomials. The sag data of the corneas was determined using the real height data measured with a Humphrey Atlas corneal topographer. The corneal topographer measures the height (z_i) at a discrete number of points. The corneal surface can then be expressed as a linear combination of the first 15 Zernike polynomials (as described in Equation 1, above), where Z_i is the ith Zernike polynomial and a_i is the weighting coefficient for this polynomial. The Zernike polynomials are a set of complete orthogon polynomials defined on a unit circle. These polynomials as listed in Table 1 above and the weighting coefficients (a_i) are calculated from the height data using a Grahm-Schmi orthogonalization procedure. The Zernike coefficients (a_i) for the 10 sample corneas are listed in Table 2 in mm.

	ACH	ASA	CGR	CNR	FCA	FCM	FCZ
a1	-7.713337	-6.698643	-7.222353	-7.169027	-7.001356	-7.322624	-7.0
a2	0.000271	-0.000985	0.000386	-0.000519	0.000426	-0.000094	-0.00
a3	0.000478	-0.000002	-0.000847	0.000996	-0.000393	0.000045	0.00
a4	0.073309	0.083878	0.077961	0.078146	0.080111	0.077789	0.07
a5	-0.000316	-0.000753	0.000119	0.000347	-0.001197	0.00022	-0.00
a6	0.001661	0.000411	-0.000148	-0.000386	0.000314	0.000669	0.0
a7	0.000193	0.00006	-0.000295	0.000324	-0.000161	-0.000058	0.00
a8	0.000098	-0.000437	0.000146	-0.00018	0.000147	0.000039	-0.00
a9	-0.000091	-0.000168	-0.000107	0.000047	-0.000181	-0.000154	-0.00
a10	-0.000055	0.000139	-0.000132	-0.000149	0.000234	-0.000228	0.00
a11	0.000277	0.000394	0.000203	0.000305	0.000285	0.000315	0.00
a12	-0.000019	-0.000105	0.000025	0.00007	-0.000058	-0.000033	0.0
a13	0.000048	0.000032	0.000085	0.000017	0.000039	0.000059	0.00
a14	-0.000067	0.000041	-0.000081	-0.000049	0.000118	-0.000108	0.00
a15	-0.000048	-0.000075	-0.000073	-0.000019	-0.000036	-0.000119	-0.00

	FGP	JAE	JBH
a1	-7.84427	-7.582005	-6.890056
a2	-0.00056	-0.000344	-0.000155
a3	0.000347	0.000246	-0.000558
a4	0.072595	0.075803	0.081415
a5	0.000686	-0.000388	-0.000269
a6	-0.00048	0.001688	0.001492
a7	0.00014	0.000104	-0.000227
a8	-0.00025	-0.000173	-0.000116
a9	0.000092	-0.000023	-0.000109
a10	-8.2E-05	-0.000004	0.000065
a11	0.000308	0.000309	0.0004
a12	-2E-06	-0.000115	-0.00011
a13	0.000101	-0.000042	-0.000052
a14	-1.9E-05	-0.000068	0.00001
a15	0.000022	-0.000013	-0.000048

Table 2: The Zernike coefficients for the 10 individual corneal surfaces in mm.

These wavefront aberration coefficients can be calculated using optical design software such as OSLO (Sinclair Optics). Table 3 shows the results of calculating the wavefront aberration for subject FCM. (N.B. The normalization factor for the polynomials used in OSLO is different from those shown in Table 3. This difference has been incorporated into the coefficient values.)

	Aberration Coefficients for FCM (OSLO)
A0	-0.000123
A1	4.5960e-07
A2	2.0869e-07
A3	-5.355e-06
A4	0.000551
A5	0.000182
A6	3.7296e-05
A7	-5.5286e-05
A8	0.000116
A9	-0.000217
A10	-0.000147
A11	-3.8151e-05
A12	6.1808e-05
A13	-3.3056e-07
A14	4.888e-07
A15	-1.8642e-06
A16	-0.000115
A17	-0.000127

Table 3: The corneal aberration coefficients in mm calculated for subject FCM using OSLO(N.B. OSLO numbering order)

Example 2

An averaged design embodiment of the inventive lenses has been calculated using the average "old" cornea information provided by Pablo Artal, Murcia, Spain. This data was taken from a population sample of 16 old corneas in which all of the subjects had a visual acuity of 20/30 or better. The corneal surfaces were described using Zernike polynomials for an aperture of 2.0 mm radius (r_o). The polynomial coefficients were then used to determine the radius and asphericity values using Equations 2 and 3.

$$R = \frac{r_o^2}{2(2\sqrt{3}a_4 - 6\sqrt{5}a_{11})} \qquad (2)$$

$$K^2 = \frac{8R^3}{r_o^4} 6\sqrt{5}a_{11} \qquad (3)$$

Note that the asphericity constant, K, describes the surface's variation from a sphere ($K^2=1-e^2$). (i.e. For a sphere K = 1 and for a parabola K = 0). (cc = K^2-1, wherein cc is the conical constant)

Because the cornea surface has only been described for a central diameter of 4 mm, the calculated R and K are also only accurate over the central 4 mm. A pupil size of 4.0 mm is therefore selected for design purposes. This pupil size is reasonable for intraocular lens design purposes.

A 22D CeeOn® 911 lens from Pharmacia Corp was chosen as a starting point for the averaged lens design. For the purpose of comparison, the averaged lenses were also designed to be 22D. (Note that other dioptres would give similar simulation results, provided that the spherical surfaces of the lenses is the same.) The surface information fo the starting point eye model is summarized in Table 4. In the conical and aspherical data demonstrated in Table 4, the average conic constant CC is determined for the 10 individual corneas of Example 1.

Surface #	Radius (mm)	Thickness (mm)	Aperture Radius (mm)	Conic Constant	Refractive Index
Object	--	∞	2.272611		1.0
1 (cornea)	7.573	3.6	2.272611	-0.0784 *	1.3375
2 (pupil)	--	--	2.0	--	1.3375
3	--	0.9	2.0	--	1.3375
4 (lens 1)	11.043	1.14	3.0	--	1.4577
5 (lens 2)	-11.043	17.2097	3.0	--	1.336

Table 4: Surface data for the starting point of the averaged ("Z11") design

* This conic constant for the "average" cornea is taken from the published works of Guirao and Artal

The wavefront aberration coefficients in mm for the average cornea are shown in column 1 of Table 5, while the coefficients in mm of the combination of the average cornea and the 911 lens are shown in column 2 of Table 5. Note that the Z11 coefficient (a11) of the average old cornea alone is 0.000220 mm, while the Z11 of this eye implanted with a 911 would be 0.000345 mm.

	Average Cornea	Average Cornea +911
a1	0.000432	0.000670
a2	0.0	0.0
a3	0.0	0.0
a4	0.000650	0.00101
a5	0.0	0.0
a6	0.0	0.0
a7	0.0	0.0
a8	0.0	0.0
a9	0.0	0.0
a10	0.0	0.0
a11	0.000220	0.000345
a12	0.0	0.0
a13	0.0	0.0
a14	0.0	0.0
a15	0.0	0.0

Table 5: Zernike coefficients in mm of the average cornea and the starting point for design (Average cornea + 911)

The averaged lens was optimized to minimize spherical aberration, while maintaining a 22D focal power. The lens material remained the same as in the 22D 911 lens (HRI silicone, the refractive index of which is 1.45 in77 at 37°C . The resulting design for an equiconvex lens is The averaged lens was optimized to minimize spherical aberration, while maintaining a 22D focal power. The lens material remained the same as in the 22D 911 lens (HRI silicone, the refractive The resulting design for an equiconvex lens is provided in Table 6. The total-eye Z11 coefficient of the average cornea combined with this lens is $-2.42\text{x } 10^{-7}$ mm(versus 0.000345 mm for the cornea plus 911 lens).

Surface #	Radius (mm)	Thickness (mm)	Aperture Radius (mm)	Conic Constant	4^{th} Order Aspheric Constant
Object	--	∞	2.272611		
1 (cornea)	7.573	3.6	2.272611	-0.0784	
2 (pupil)	--	--	2.0	--	
3	--	0.9	2.0	--	
4 (lens 1)	10.0	1.02	3.0	-2.809	-0.000762
5 (lens 2)	-12.0	17.2097	3.0	--	

Surface #	6^{th} Order Aspheric Constant	Refractive Index
Object		1.0
1 (cornea)		1.3375
2 (pupil)		1.3375
3		1.3375
4 (lens 1)	-1.805e-05	1.4577
5 (lens 2)		1.336

Table 6: Surface data for the starting point of the averaged lens design

The corneas of the 10 test subjects were combined in an optical system with both the 911 and the averaged lenses. The resulting total-eye Z11 coefficients are shown in Fig. 1. As demonstrated, in Fig. 1, in each case, the absolute value of the Z11 coefficient was less when the Z11 lens was implanted. Because subjects CGR and FCZ have relatively low levels of corneal spherical aberration to begin with, the total-eye spherical aberration is overcorrected in these two cases. As a result, the sign of the total spherical aberration is noticeably reversed in these two cases, and the amount of spherical aberration is still significant. In every other case, the spherical aberration of the total eye would be essentially 0 after the implantation of a Z11 lens. The visual acuity of each of the 10 test subjects were calculated according to standard methods described in "Visual acuity modeling using optical raytracing of schematic eyes", Greivenkamp et al., American journal of ophthalmology, 120(2), 227-240, (1995), for the implantation of both a 22D 911 lens and a 22D averaged "Z11" lens. The square wave responses were calculated using OSLO™ and a software module was written in Matlab™ to calculate the visual acuity following the above method. The resulting visual acuities are shown in Fig.

2. Out of the 10 cases investigated and shown in Fig. 2, eight subjects had better vision when implanted with the averaged lens according to the present invention. In the cases where the visual acuity decreased their Snellen distance increased by less than 1 ft whicl would not show up in visual acuity testing.

To be able to assess the optical quality difference between a CeeOn® 911A and averaged lenses according to the present invention, a physical model of the average cornea was designed and manufactured. It is a convex-plano lens of PMMA with an aspheric front surface having a value of 0.000218 for Zernike coefficient a11. This valu€ is almost equal to the value of the calculated average cornea: 0.000220. With the PMM/ model cornea MTF measurements were performed on an optical bench in a model eye with the "averaged" Z11 lenses and CeeOn® 911A lenses. Modulation Transfer Functio (MTF) measurements are a widely accepted method of quantifying image quality. Through focus MTF measurements at 50c/mm and a frequency MTF curves focussed at 50c/mm, in both cases with a 3mm pupil are shown in the Fig. 3 and Fig. 4, respectively for lenses with an optical power of 20D. The width of the through focus MTF at 0.2 MT units is a measure for the depth of focus and is equal for both lenses. The MTF curve focussed at 50c/mm for "averaged" Z11 lenses is almost diffraction limited and is better than that for CeeOn 911A lenses.

The astigmatism of the cornea and the defocus of the system can be corrected by adjusting the Zernike coefficients of the cornea model and the focal position of the system. When this is done and the procedure for calculating visual acuity is repeated the results in Fig. 6 are obtained. They represent a modeled best corrected visual acuity. We now see that, in all cases, after correction for astigmatism and defocus (as in reality would be done with spectacles) the inventive averaged lens produces a higher best corrected visual acuity than the 911 lens of the same dioptre.

Example 3

Individually Designed Lenses:

As a potential further improvement upon the averaged lens ("Z11 lenses"), an individualized lens ("I11 lenses") was designed for each of four subject corneas employing the same design principles as demonstrated in Example 2. The individual lenses were designed such that the Z11 of the lens balances the Z11 of the individual cornea. The total-eye Z11 coefficients for the I11 lenses are shown in Table 7, together

Subject	911	averaged	individual
CGR	0.000107	-0.000101	-0.000004
FCZ	0.000105	-0.000100	-0.000005
JAE	0.000195	-0.000016	-0.000012
JBH	0.000238	0.000037	-0.000019

Table 7: The Z11 coefficients in mm of the model eyes with the 911, Z11 and I11 lenses

with the corresponding coefficients for the 911 and the averaged lenses. Furthermore, for each of the 911, Z11 (averaged), and I11 (individual) lenses, the MTF curve at best focussed at 50 c/mm and the through focus MTF at 50c/mm for subject JAE are plotted below in Fig. 7 and 8. From Fig. 7 and 8, it is seen that the MTF at 50 c/mm of eyes implanted with the Z11 and I11 lenses is higher than the MTF of the same eyes implantec with 911 lenses. It can also be seen that the through focus MTF of all of the lenses is satisfactory. The Z11 has as much depth of focus as the 911. However, it is also interesting to note that the I11 does not provide a significant improvement in either MTF or through focus MTF, relative to the Z11 lens.

The visual acuities of the subjects with individualized lenses have also been calculated. Fig. 9 compares these acuities with the visual acuity calculated for the 911 and Z11 lenses.

From Fig. 9, we see that for all 4 subjects, visual acuity is better for both the Z11 and I11 lenses than it is for the 911 lens. We also see that the results with the Z11 and I11 lenses

do not differ significantly - the average cornea is relatively accurate for each of the 4 tes subjects.

Example 4

The design of an ophthalmic lens, which is adapted to reduce the spherical aberration of an average cornea obtained from a group of people will be described in detail here below. The lens will be called the Z11 lens because it compensates for the normalized 11^{th} Zernike term describing spherical aberration of the corneas. It was decided to use a population of potential recipients of the Z11 lens, namely cataract patients.

Description of the Population:

The population included 71 cataract patients from St. Erik's eye hospital in Stockholm, Sweden. These patients were of ages ranging from 35 to 94 years (as of Apr 12, 2000). The average age of our population was 73.67 years. A histogram of the age c the population is shown in Figure 10.

The corneas of the 71 subjects were measured using an Orbscan® (Orbtek, Salt Lake City) device. Orbscan® is a scanning slit -based, corneal and anterior segment topographer that measures both surfaces of the cornea, as well as the anterior lens surfac and the iris. Each surface can be displayed as maps of elevation, inclination, curvature, and power.

Fitting Algorithm:

The corneal elevation height data (the Cartesian locations of points on the surfac of the cornea) for the anterior surface was obtained using the Orbscan®, and used as rav data for the determination of the optical properties of the cornea. The height data from a example Orbscan® file is represented in Figure 11.

The Cartesian co-ordinates representing the elevation height data are transformed to polar co-ordinates (x,y,z → r,θ,z). In order to describe the surface, this data is then fit to a series of polynomials as described in Equation 1b. The coefficients (a's), or weighting factors, for each polynomial are determined by the fitting procedure resulting in a complete description of the surface. The polynomials (Z_i) used are the normalized Zernike polynomials.

$$z(\rho,\theta) = \sum_{i=1}^{L} a_i Z_i \qquad (1b)$$

These polynomials are special because they are orthonormal over a continuous unit circle. They are commonly used to describe wavefront aberrations in the field of optics. Corneal topographers measure the elevation heights at a discrete number of points The Zernike polynomials are not orthogonal over a discrete set of points. However, applying an orthogonalization procedure, termed Gram-Schmidt orthogonalization, to the height data, allows the data to be fit in terms of Zernike polynomials maintaining the advantages of an orthogonal fit. Sixty-six coefficients (a's) were used to fit the height dat provided by the Orbscan® software. A Matlab™ algorithm was used in the fitting procedure. The radius and asphericity value can be approximated from the Zernike coefficients (Equations 2b and 3b) and the conic constant of the surface is simply K^2-1 (from this we know that for a sphere $K^2 = 1$). The fitting procedure is well described in a number of references. Four different articles are refereed to here: "Wavefront fitting with discrete orthogonal polynomials in a unit radius circle", Daniel Malacara, Optical Engineering, June 1990, Vol. 29 No. 6, "Representation of videokeratoscopic height data with Zernike polynomials", J. Schwiegerling, J.Greivenkamp and J.Miller, JOSA A, October 1995, Vol. 12 No. 10, "Wavefront interpretation with Zernike polynomials" J.W.Wang and D.E. Silva, Applied Optics, May 1980, Vol. 19, No. 9 and "Corneal wave aberration from videokeratography: accuracy and limitations of the procedure", Antonio Guirao and Pablo Artal, J Opt Soc Am A Opt Image Sci Vis Jun 2000, Vol. 17(6):955-65

$$R = \frac{r_{pupil}^2}{2(2\sqrt{3}a_4 - 6\sqrt{5}a_{11})} \qquad (2b)$$

$$K_{sq} = \frac{8R^3}{r_o^4} 6\sqrt{5} a_{11} \qquad (3b)$$

Calculation of Wavefront Aberration:

Knowing the shape of the anterior corneal surface (Zernike coefficients describe above as a's), it is possible to determine the wavefront aberration contributed by this surface using a raytrace procedure. This is described in for example "Corneal wave aberration from videokeratography: accuracy and limitations of the procedure", Antonio Guirao and Pablo Artal, J Opt Soc Am A Opt Image Sci Vis Jun 2000, Vol. 17(6):955-6

Results:

Average Corneal Spherical Aberration and Shape:

The 71 subjects were evaluated using the criteria described above for a 6 mm aperture. The wavefront aberration of each subject was determined after fitting the surface elevation with Zernike polynomials. Figure 12 shows that average and standard deviation of each Zernike term (normalized format). The error bars represent ± 1 standar deviation. There are three aberrations that are significantly different from zero on averag in our population. These are astigmatism (A5), coma (A9) and spherical aberration (A11 Spherical aberration is the only rotationally symmetric aberration, making it the only aberration that can be corrected with a rotationally symmetric IOL.

Figure 13 shows a scatter plot of the value of the Zernike coefficient (OSLO format) representing spherical aberration for each of the 71 subjects before cataract surgery. The solid line in the middle represents the average spherical aberration, while th dotted lines represent +1 and −1 standard deviation. Table 8 lists the average, standard deviation, maximum and minimum values for the radius, aspheric constant, spherical aberration and root mean square error.

	Average Value	Standard Deviation	Maximum	Minimum
R (mm)	7.575	0.333	8.710	7.072
Ksq	0.927	0.407	2.563	0.0152
SA coefficient OSLO format (in mm)	0.000604	0.000448	0.002003	-0.000616
RMSE	0.000055	0.00000482	0.000069	0.000045

Table 8: the average, standard deviation, maximum and minimum values for the radius, aspheric constant, spherical aberration and root mean square error for a 6mm aperture.

Tables 9,10 and 11 below show the corresponding results for aperture sizes of 4,5 and 7 mm respectively. Figure 14,15 and 16 are the corresponding scatter plots.

	Average Value	Standard Deviation	Maximum	Minimum
R	7.56292	0.320526	8.688542	7.067694
Ksq	0.988208	0.437429	2.33501	-0.051091
SA (A11 in mm)	0.000139	0.000103	0.00041	-0.000141
RMSE	4.52E-05	4E-06	0.000054	0.000036

Table 9: The average, standard deviation, maximum and minimum values for the radius, aspheric constant, spherical aberration and root mean square error using an aperture diameter of 4 mm.

	Average Value	Standard Deviation	Maximum	Minim
R	7.55263	0.320447	8.714704	7.0909
Ksq	0.945693	0.364066	2.045412	0.0446
SA (A11 in mm)	0.00031189	0.000208	0.000793	-0.0002
RMSE	4.7E-05	4.02E-06	0.000057	0.0000

Table 10: The average, standard deviation, maximum and minimum values for the radius aspheric constant, spherical aberration and root mean square error using an aperture diameter of 5 mm.

	Average Value	Standard Deviation	Maximum	Minimum
R	7.550226	0.336632	8.679712	7.040997
Ksq	0.898344	0.416806	2.655164	-0.04731
SA (A11 in mm)	0.001058	0.000864	0.003847	-0.001319
RMSE	7.58E-05	1.02E-05	0.000112	0.000057

Table 11: The average, standard deviation, maximum and minimum values for the radius aspheric constant, spherical aberration and root mean square error using an aperture diameter of 7 mm.

Design Cornea:

One model cornea was designed and each Z11 lens power was designed using thi cornea. The cornea was designed so that it had a spherical aberration that is the same as the average calculated for the population. The design cornea radii and aspheric constants are listed in Table 12 for different aperture sizes. In every case, the radius of curvature was taken to be the average radius determined from the Zernike fit data. The aspheric constant was varied until the spherical aberration value of the model cornea was equal to the average spherical aberration value for the population.

Aperture size (mm)	Radius (mm)	Conic Constant (OSLO value, K^2-1)	Z11 Coefficient (mm)
4	7.563	-0.0505	0.000139
5	7.553	-0.1034	0.000312
6	7.575	-0.14135	0.000604
7	7.55	-0.1810	0.001058

Table 12: The design cornea radii and aspheric constants for aperture diameters of 4, 5, 6 and 7 mm.

As discussed previously, the 6 mm aperture diameter values are used for the design cornea. This choice enables us to design the Z11 lens so that it has no spherical aberration (when measured in a system with this cornea) over a 5.1 mm lens diameter. The OSLO surface listing for the Z11 design cornea is listed in Table 13. The refractive index of the cornea is the keratometry index of 1.3375.

These values were calculated for a model cornea having a single surface with a refractive index of the cornea of 1.3375. It is possible to use optically equivalent model formats of the cornea without departing from the scope of the invention. For example multiple surface corneas or corneas with different refractive indices could be used.

Surface #	Radius (mm)	Thickness (mm)	Aperture Radius (mm)	Conic Constant (cc)	Refractive index
Object	--	1.0000e+20	1.0000e+14	--	1.0
1 (cornea)	7.575000	3.600000	3.000003	-0.141350	1.3375
2 (pupil)	--	--	2.640233	--	1.3375
3	--	0.900000	2.64023	--	1.3375
4		25.519444	2.550292	--	1.3375
5			2.2444e-05	--	1.3375

Table 13: OSLO surface listing for the Z11 design cornea.

Lens Design:

Each Z11 lens was designed to balance the spherical aberration of the design cornea. The starting point for design was the CeeOn Edge© 911 lens described in U.S. 5,444,106 of the same power, with modified edge and center thickness. The lens was the placed 4.5 mm from the anterior corneal surface. The distance from the anterior corneal surface is not that critical and could be varied within reasonable limits. The surface information for the starting point eye model for the 22 D lens design process is listed in Table 14. The anterior surface of the lens was described using the formula shown in Equation 4. The variables **cc, ad** and **ae** were modified to minimize the spherical aberration. The variables are determined for an aperture size of 5.1 mm and the surface i extrapolated from these values to the optical aperture size of 6 mm. The resulting 22D Z11 eye model is listed in Table 15. The anterior surface of this 22D lens has been modified in such a way that the spherical aberration of the system (cornea+lens) is now approximately equal to 0. The wavefront aberration coefficients as calculated by OSLO for the CeeOn Edge 911 22D lens eye model and the 22D Z11 lens eye model are listed below in Table 16. Note that the coefficient representing spherical aberration for the starting point eye model is 0.001005 mm for a 6mm diameter aperture placed at the cornea, while the same coefficient for the eye model with the designed Z11 lens is -1.3399e-06 mm. The same process as described above for a 22D lens can similarly be performed for any other lens power.

$$z = \frac{(1/R)r^2}{1+\sqrt{1-(\frac{1}{R})^2(\mathbf{cc}+1)r^2}} + \mathbf{ad}r^4 + \mathbf{ae}r^6 \qquad (4)$$

Surface #	Radius (mm)	Thickness (mm)	Aperture Radius (mm)	Conic Constant (cc)	Refractive index
Object	--	1.0000e+20	1.0000e+14	--	1.0
1 (cornea)	7.575	3.600000	3.000003	-0.14135	1.3375
2 (pupil)	--	--	2.640233	--	1.336
3	--	0.900000	2.64023	--	1.336
4 (lens)	11.043	1.164	2.550191	--	1.458
5 (lens)	-11.043	17.1512	2.420989	--	1.336
6 (image)	0.0	-0.417847	0.058997	--	--

Table 14: Surface data for the starting point averaged eye model and a 22D lens

Surface #	Radius (mm)	Thickness (mm)	Aperture Radius (mm)	Conic Constant (cc)	4^{th} Order aspheric Constant
Object	--	1.0e+20	1.00e+14	--	
1 (cornea)	7.575	3.60	3.00	-0.14135	
2 (pupil)	--	--	2.64	--	
3	--	0.90	2.64	--	
4 (lens)	11.043	1.164	2.55	-1.03613	-0.000944
5 (lens)	-11.043	17.1512	2.42	--	
6 (image)	--	--	1.59e-05	--	--

Surface #	6th Order aspheric constant	Refractive index
Object		1.0
1 (cornea)		1.3375
2 (pupil)		1.336
3		1.336
4 (lens)	-1.37e-05	1.458
5 (lens)		1.336
6 (image)	--	--

Table 15: Surface data for the averaged eye model and the final 22D Z11 lens

Coefficient	Average cornea+ 22D 911	Average cornea + 22D Z11
a1	-0.000962	-1.896e-06
a2	0.0	0.0
a3	0.0	0.0
a4	2.3101e-05	-3.9504e-06
a5	0.0	0.0
a6	0.0	0.0
a7	0.0	0.0
a8	0.0	0.0
a9	0.00105	-1.3399e-06
a10	0.0	0.0
a11	0.0	0.0
a12	0.0	0.0
a13	0.0	0.0
a14	0.0	0.0
a15	0.0	0.0

Table 16: Zernike coefficients (OSLO format) for the average cornea and a 22D 911 lens and the average cornea and the 22D Z11 lens

The optical form chosen for the new Z11 design is an equiconvex lens made from a silicone with refractive index of 1.458. The spherical aberration of an average cornea is balanced by the Z11 lens yielding a system without spherical aberration. The front surface of the lens is modified such that the optical path lengths of all on-axis rays within the design aperture are the same producing a point focus. This feature can be achieved with many lens forms. The Z11 lens could therefore be designed on a convex-plano, plano-convex, non-equiconvex lens or any other design yielding a positive lens. The Z11 concept could also be extended in order to encompass a negative lens used to correct the refractive errors of the eye. The front surface or back surface could also be modified to produce the needed change in optical path difference that neutralizes the spherical

aberration. There are therefore many possible designs that would achieve the goals of th Z11 lens design.

Claims

1. A method of designing an intraocular lens capable of reducing aberrations of the eye after its implantation, comprising the steps of:

   (i) characterizing at least one corneal surface as a mathematical model;
   (ii) calculating the resulting aberrations of said corneal surface(s) by employing said mathematical model;
   (iii) selecting the optical power of the intraocular lens;
   (iv) modeling an intraocular lens such that a wavefront arriving from an optical system comprising said lens and corneal model obtains reduced aberrations.

2. A method according to claim 1, comprising determining the resulting aberrations of said corneal surface(s) in a wavefront having passed said cornea.

3. A method according to claim 1, wherein said corneal surface(s) is(are) characterized in terms of a conoid of rotation.

4. A method according to claim 1 wherein said corneal surface(s) is(are) characterized in terms of polynomials.

5. A method according to claim 4, wherein said corneal surface(s) is(are) characterized in terms of a linear combination of polynomials.

6. A method according to claim 1, wherein said optical system further comprises complementary means for optical correction, such as spectacles or an ophthalmic correction lens.

7. A method according to claim 1, wherein estimations of corneal refractive power and axial eye length designate the selection of lens optical power.

8. A method according to claim 4, wherein an optical system comprising said corneal model and modeled intraocular lens provides for a wavefront substantially reduced from aberrations as expressed by at least one of said polynomials.

9. A method according to claim 1, wherein modeling the intraocular lens includes selecting the anterior radius and surface of the lens, the posterior radius and surface of th lens, lens thickness and refractive index of the lens.

10. A method according to claim 9, wherein the aspheric component of the anterior surface is selected while the model lens has predetermined central radii, lens thickness and refractive index.

11. A method according to claim 1 including characterizing the front corneal surface of an individual by means of topographical measurements and expressing the corneal aberrations as a combination of polynomials.

12. A method according to claim 11 including characterizing front and rear corneal surfaces of an individual by means of topographical measurements and expressing the total corneal aberrations as a combination of polynomials.

13. A method according to claim 1, including characterizing corneal surfaces of a selected population and expressing average corneal aberrations of said population as a combination of polynomials.

14. A method according to claim 1, comprising the further steps of :

(v) calculating the aberrations resulting from an optical system comprising said modeled intraocular lens and cornea;

(vi) determining if the modeled intraocular lens has provided sufficient reductior in aberrations; and optionally re-modeling the intraocular lens until a sufficient reduction is obtained.

15. A method according to claim 14, wherein said aberrations are expressed as a linear combination of polynomials.

16. A method according to claim 15, wherein the re-modeling includes modifying one or several of the anterior surface and curvature, the posterior radius and surface, lens thickness and refractive index of the lens.

17. A method according to claim 4 or 5, wherein said polynomials are Seidel or Zernike polynomials.

18. A method according to claim 17, comprising the steps of:

(i) expressing the corneal aberrations as a linear combination of Zernike polynomials;

(ii) determining the corneal wavefront Zernike coefficients;

(iii) modeling the intraocular lens such that an optical system comprising sai model lens and cornea provides a wavefront having a sufficient reductio of Zernike coefficients.

19. A method according to claim 18, further comprising the steps of :

(iv) calculating the Zernike coefficients of a wavefront resulting from an optical system comprising the modeled intraocular lens and cornea;

(v) determining if said intraocular lens has provided a sufficient reduction o Zernike coefficients; and optionally re-modeling said lens until a sufficient reduction is said coefficients are obtained.

20. A method according to claim 19, comprising sufficiently reducing Zernike coefficients referring to spherical aberration.

21. A method according to claim 19 comprising sufficiently reducing Zernike coefficien referring to aberrations above the fourth order.

22. A method according to claim 20 comprising sufficiently reducing the 11th Zernike coefficient of a wavefront front from an optical system comprising cornea and said modeled intraocular lens, so as to obtain an eye sufficiently free from spherical aberration.

23. A method according to claim 19, wherein the re-modeling includes modifying one o: several of the anterior radius and surface, the posterior radius and surface, lens thickness and refractive index of the lens.

24. A method according to claim 23, comprising modifying the anterior surface of the lens until a sufficient reduction in aberrations is obtained.

25. A method according to claim 17, comprising modeling a lens such that an optical system comprising said model of intraocular lens and cornea provides reduction of spherical and cylindrical aberration terms as expressed in Seidel or Zernike polynomials in a wave front having passed through the system.

26. A method according to claim 25, obtaining a reduction in higher aberration terms.

27. A method according to claim 8 comprising:

(i) characterizing corneal surfaces of a selected population and expressing each cornea as a linear combination of polynomials;
(ii) comparing polynomial coefficients between individual corneas;
(iii) selecting one nominal coefficient value from an individual cornea;
(iv) modeling a lens such that a wavefront resulting arriving from an optical system comprising said lens and individual cornea sufficiently reduces said nominal coefficient value.

28. A method according to claim 27, wherein said polynomial coefficient refers to the Zernike aberration term expressing spherical aberration.

29. A method according to claim 27,wherein said nominal coefficient value is the lowest within the selected population.

30. A method of selecting an intraocular lens that is capable of reducing aberrations of the eye after its implantation comprising the steps of:

(i) characterizing at least one corneal surface as a mathematical model;
(ii) calculating the resulting aberrations of said corneal surfaces(s) by employing said mathematical model;
(iii) selecting an intraocular lens having a suitable optical power from a plurality of lenses having the same power, but different aberrations;
(iv) determining if an optical system comprising said selected lens and corneal model sufficiently reduces the aberrations.

31. A method according to claim 30, comprising determining the resulting aberrations of said corneal surface(s) in a wavefront having passed said cornea.

32. A method according to claim 30 further comprising the steps of:

(v) calculating the aberrations of a wave front arriving from an optical system of said selected lens and corneal model;
(vi) determining if said selected intraocular lens has provided a sufficient reduction in aberrations in a wavefront arriving from said optical system; and optionally repeating steps (iii) and (iv) by selecting at least one new lens having the same optical power until finding a lens capable of sufficiently reducing the aberrations.

33. A method according to claim 30, wherein said corneal surface(s) is(are) characterized in terms of a conoid of rotation.

34. A method according to claim 30 wherein said corneal surface(s) is(are) characterized in terms of polynomials.

35. A method according to claim 34, wherein said corneal surface(s) is(are) characterized in terms of a linear combination of polynomials.

36. A method according to claim 30 or 32, wherein said optical system further comprises complementary means for optical correction, such as spectacles or an ophthalmic correction lens.

37. A method according to claim 30, wherein corneal refractive power and axial eye length estimations designate the selection of lens optical power.

38. A method according to claim 34 or 35, wherein an optical system comprising said corneal model and selected intraocular lens provides for a wavefront substantially reduced from aberrations as expressed by at least one of said polynomials.

39. A method according to claim 30 including characterizing the front corneal surface of an individual by means of topographical measurements and expressing the corneal aberrations as a combination of polynomials.

40. A method according to claim 39 including characterizing front and rear corneal surfaces of an individual by means of topographical measurements and expressing the total corneal aberrations as a combination of polynomials.

41. A method according to claim 30, including characterizing corneal surfaces of a selected population and expressing average corneal aberrations of said population as a combination of polynomials.

42. A method according to claim 38, wherein said polynomials are Seidel or Zernike polynomials.

43. A method according to claim 42, comprising the steps of:

(i) expressing the corneal aberrations as a linear combination of Zernike polynomials;

(ii) determining the corneal Zernike coefficients;

(iii) selecting the intraocular lens such that an optical system comprising said lens and cornea provides a wavefront having a sufficient reduction in Zernike coefficients.

44. A method according to claim 43, further comprising the steps of :

(iv) calculating the Zernike coefficients resulting from an optical system comprising the modeled intraocular lens and cornea;

(v) determining if said intraocular lens has provided a reduction of Zernike coefficients; and optionally selecting a new lens until a sufficient reduction is said coefficients is obtained.

45. A method according to claim 43 or 44, comprising determining Zernike polynomials up to the 4th order.

46. A method according to claim 45 comprising sufficiently reducing Zernike coefficient referring to spherical aberration.

47. A method according to claim 46 comprising sufficiently reducing Zernike coefficient above the fourth order.

48. A method according to claim 46 comprising sufficiently reducing the 11th Zernike coefficient of a wavefront front from an optical system comprising model cornea and said selected intraocular lens, so as to obtain an eye sufficiently free from spherical aberration

49. A method according to claim 39 comprising selecting a intraocular lens such that an optical system comprising said intraocular lens and cornea provides reduction of spherical aberration terms as expressed in Seidel or Zernike polynomials in a wave fron having passed through the system.

50. A method according to claim 39, wherein reduction in higher aberration terms is accomplished.

51. A method according to claim 30 characterized by selecting an intraocular lens from kit comprising lenses with a suitable power range and within each power range a plurali of lenses having different aberrations.

52. A method according to claim 51, wherein said aberrations are spherical aberrations.

53. A method according to claim 51, wherein said lenses within each power range have surfaces with different aspheric components.

54. A method according to claim 53, wherein said surfaces are the anterior surfaces.

55. A method of designing an ophthalmic lens suitable for implantation into the eye, characterized by the steps of:

- selecting a representative group of patients;
- collecting corneal topographic data for each subject in the group;
- transferring said data to terms representing the corneal surface shape of each subject for a preset aperture size;
- calculating a mean value of at least one corneal surface shape term of said group, so as to obtain at least one mean corneal surface shape term and/or calculating a mean value of at least one to the cornea corresponding corneal wavefront aberration term, each corneal wavefront aberration term being obtained by transformation through corneal surface shape terms;

- from said at least one mean corneal surface shape term or from said at least one mea corneal wavefront aberration term designing an ophthalmic lens capable of reducing said at least one mean wavefront aberration term of the optical system comprising cornea and lens.

56. Method according to claim 55, characterized in that it further comprises the steps of:
- designing an average corneal model for the group of people from the calculated at least one mean corneal surface shape term or from the at least one mean corneal wavefront aberration term;
- checking that the designed ophthalmic lens compensates correctly for the at least one mean aberration term by measuring these specific aberration terms of a wavefront having traveled through the model average cornea and the lens and redesigning the lens if said at least one aberration term not has been sufficiently reduced in the measured wavefront.

57. Method according to claim 55 or 56, characterized by calculating surface descriptive constant for the lens to be designed from the mean corneal surface shape terms or from the mean corneal wavefront aberration terms for a predetermined radius.

58. Method according to any one of the claims 55-57, characterized by selecting people in a specific age interval to constitute the group of people.

59. Method according to any one of the claims 55-58, characterized by selecting people who will undergo cataract surgery to constitute the group of people.

60. Method according to any one of the claims 55-59, characterized by designing the len specifically for a patient that has undergone corneal surgery and therefor selecting peopl who have undergone corneal surgery to constitute the group of people.

61. Method according to any one of the claims 55-60, characterized by selecting people who have a specific ocular disease to constitute the group of people.

62. Method according to any one of the claims 55-61, characterized by selecting people who have a specific ocular optical defect to constitute the group of people.

63. Method according to any one of the claims 55-62, characterized in that it further comprises the steps of:

- measuring the at least one wavefront aberration term of one specific patient's cornea
- determining if the selected group corresponding to this patient is representative for this specific patient and if this is the case implant the lens designed from these average values and if this not is the case implant a lens designed from average values from another group or design an individual lens for this patient.

64. Method according to any one of the claims 55-63, characterized by providing the len with at least one nonspheric surface that reduces at least one aberration term of an incoming nonspheric wavefront.

65. Method according to claim 64, characterized in that said aberration term is a positive spherical aberration term.

66. Method according to any one of the claims 55-65, characterized by providing the len with at least one nonspheric surface that reduces at least one term of a Zernike polynomial representing the aberration of an incoming nonspheric wavefront.

67. Method according to claim 66, characterized by providing the lens with at least one nonspheric surface that reduces the 11^{th} normalized Zernike term representing the spherical aberration of an incoming nonspheric wavefront.

68. A method according to any of claims 55-67 characterized by designing a lens to reduce spherical aberration in a wavefront arriving from an average corneal surface having the formula: $z = \frac{(1/R)r^2}{1+\sqrt{1-(\frac{1}{R})^2(cc+1)r^2}} + adr^4 + aer^6$

wherein the conical constant cc has a value ranging between −1 and 0, R is the centra lens radius and ad and ae are aspheric constants.

69. A method according to claim 68, wherein the conical constant (cc) ranges from abou −0.05 for an aperture size (pupillary diameter) of 4 mm to about −0.18 for an apertur size of 7 mm.

70. Method according to claim 68, characterized by providing the lens with a surface described by a modified conoid having a conical constant (cc) less than 0.

71. Method according to any one of the claims 55-70, characterized by providing the lens with a, for the patient, suitable refractive power, this determining the radius of the lens.

72. Method according to any one of the claims 55-71, characterized by designing the lens to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration with a value in the interval from 0.000156 mm to 0.001948 mm for a 3 mm aperture radius using polynomials expressed in OSLO format.

73. Method according to any one of the claims 55-71, characterized by designing the lens to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration with a value in the interval from 0.000036 mm to 0.000448 mm for a 2 mm aperture radius using polynomials expressed in OSLO format.

74. Method according to any one of the claims 55-71, characterized by designing the ler to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration with a value in the interval from 0.0001039 mm to 0.0009359 mm for a 2,5 mm aperture radius using polynomials expressed in OSLO format.

75. Method according to any one of the claims 55-71, characterized by designing the ler to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration with a value in the interval from 0.000194 mm to 0.00365 mm for a 3,5 mm aperture radius using polynomials expressed in OSLO format.

76 An ophthalmic lens obtained in accordance with any of claims 1 to 75, capable of transferring a wavefront having passed through the cornea of the eye into a substantially spherical wavefront having its center in the retina of the eye.

77. An ophthalmic lens capable of compensating for the aberrations of a corneal model designed from a suitable population, such that a wavefront arriving from an optical system comprising said model cornea and said lens obtains substantially reduced aberrations.

78. An ophthalmic lens according to claim 77, wherein said corneal model includes average aberration terms calculated from characterizing individual corneas and expressing them in mathematical terms so as to obtain individual aberration terms..

79. An ophthalmic lens according to claim 77, wherein said aberration terms is a linear combination of Zernike polynomials.

80. An ophthalmic lens according to claim 79 capable of reducing aberration terms expressed in Zernike polynomials of said corneal model, such that a wavefront arriving

from an optical system comprising said model cornea and said lens obtains substantially reduced spherical aberration.

81. An ophthalmic lens according to claim 80 capable of reducing the 11th Zernike term of the 4th order.

82. An ophthalmic lens according to claim 77 being an intraocular lens.

83. An ophthalmic lens according to claim 77 adapted to replace the natural lens in a patient's eye, said ophthalmic lens having at least one nonspheric surface, this at least on nonspheric surface being designed such that the lens, in the context of the eye, provides to a passing wavefront at least one wavefront aberration term having substantially the same value but with opposite sign to a mean value of the same aberration term obtained from corneal measurements of a selected group of people, to which said patient is categorized, such that a wavefront arriving from the cornea of the patient's eye obtains a reduction in said at least one aberration term provided by the cornea after passing said lens.

84. An ophthalmic lens according to claim 83, characterized in that the surface of the len is designed to reduce at least one positive aberration term of a passing wavefront.

85 An ophthalmic lens according to claim 83 or 84, characterized in that the at least one wavefront aberration term provided to the passing wavefront by the lens is a spherical aberration term, such that a wavefront arriving from the cornea of the patient's eye obtains a reduction in said spherical aberration term provided by the cornea after passing said lens.

86. An ophthalmic lens according to any one of the claims 83-85, characterized in that th at least one wavefront aberration term provided to the passing wavefront by the lens is at least one term of a Zernike polynomial representing the wavefront aberration of the cornea.

87. An ophthalmic lens according to claim 86, characterized in that the at least one wavefront aberration term provided to the passing wavefront by the lens is the 11th normalized Zernike term of a wavefront aberration of the cornea.

88. An ophthalmic lens according to any one of the claims 83-87, characterized in that said selected group of people is a group of people belonging to a specific age interval.

89. An ophthalmic lens according to any one of the claims 83-88, characterized in that tł lens is adapted to be used by a patient that has undergone corneal surgery and in that saic selected group of people is a group of people who have undergone corneal surgery.

90. An ophthalmic lens according to any one of the claims 83-88, characterized in that said selected group of people is a group of people who will undergo a cataract surgical operation.

91. An ophthalmic lens according to claim 90 characterized in that the nonspheric surfac is a modified conoid surface having a conical constant (cc) less than zero.

92. An ophthalmic lens according to claim 91 characterized in that that is capable of eliminating or substantially reducing spherical aberration of a wavefront in the eye or in an eye model arriving from a prolate surface having the formula:

$$z = \frac{(1/R)r^2}{1+\sqrt{1-(\frac{1}{R})^2(\mathrm{cc}+1)r^2}} + \mathrm{ad}r^4 + \mathrm{ae}r^6$$

the conical constant cc has a value ranging between −1 and 0, R is the central lens radius and ad and ae are aspheric constants.

93. An ophthalmic lens according to any one of the claims 83-92, characterized in that tb lens is provided with a, for the patient, suitable refractive power less than or equal to 30 diopters.

94. An ophthalmic lens according to any one of the claims 83-93, characterized in that one of the at least one nonspheric surface of the lens is the anterior surface.

95. An ophthalmic lens according to any one of the claims 83-94, characterized in that one of the at least one nonspheric surface of the lens is the posterior surface.

96. An ophthalmic lens according to any one of the claims 83-95, characterized in that the lens is made from a soft biocompatible material.

97. An ophthalmic lens according to any one of the claims 83-96, characterized in that the lens is made of a silicone material.

98. An ophthalmic lens according to claim 97, characterized in that the silicone material is characterized by a refractive index larger than or equal to 1.43 at a wavelength of 546 nm, an elongation of at least 350 %, a tensile strength of at least 300 psi and a shore hardness of about 30 as measured with a Shore Type A Durometer.

99. An ophthalmic lens according to any one of the claims 83-98, characterized in that the lens is made of hydrogel.

100. An ophthalmic lens according to any one of the claims 83-95, characterized in that the lens is made of a rigid biocompatible material.

101. An ophthalmic lens according to any one of the claims 83-100, characterized in that it is designed to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration with a value in the interval from 0.000156 mm to 0.001948 mm for a 3 mm aperture radius using polynomials expressed in OSLO format.

102. An ophthalmic lens according to any one of the claims 83-100, characterized in tha it is designed to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration witl a value in the interval from 0.000036 mm to 0.000448 mm for a 2 mm aperture radius using polynomials expressed in OSLO format.

103. An ophthalmic lens according to any one of the claims 83-100, characterized in tha it is designed to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration witl a value in the interval from 0.0001039 mm to 0.0009359 mm for a 2.5 mm aperture radius using polynomials expressed in OSLO format.

104. An ophthalmic lens according to any one of the claims 83-100, characterized in that it is designed to balance the spherical aberration of a cornea that has a Zernike polynomial coefficient representing spherical aberration of the wavefront aberration with a value in the interval from 0.000194 mm to 0.00365 mm for a 3.5 mm aperture radius using polynomials expressed in OSLO format.

105. An ophthalmic lens having at least one nonspherical surface which when expressed as a linear combination of polynomial terms representing its aberrations is capable of reducing similar such aberration terms obtained in a wavefront having passed the cornea, thereby obtaining an eye sufficiently free from aberrations.

106. A lens according to claim 105, wherein said nonspherical surface is the anterior surface of the lens.

107. A lens according to claim 106, wherein said nonspherical surface is the posterior surface of the lens.

108. A lens according to claim 105, being an intraocular lens.

109. A lens according to claim 105, wherein said polynomial terms are Zernike polynomials.

110. A lens according to claim 109 capable of reducing polynomial terms representing spherical aberrations and astigmatism.

111. A lens according to claim 110, capable of reducing the 11th Zernike polynomial term of the 4th order.

112. A lens according to claim 105 made from a soft biocompatible material.

113. A lens according to claim 105 made of silicone.

114. A lens according to claim 105 made of hydrogel.

115. A lens according to claim 105 made of a rigid biocompatible material.

116. A method of performing visual correction in a patient by implanting an intraocular lens, which at least partly compensates for the aberrations of the eye, comprising the steps of:

- removing the natural lens from the eye;
- measuring the aberrations of the eye not comprising the lens by using a wavefront sensor;
- providing a lens that is capable reducing at least one aberration term as found by the wavefront sensing;
- implanting the lens into the eye of the patient.

117. A method according to claim 116, wherein the lens is provided by selection from a kit of lenses which includes a plurality of lenses with different capacity to correct said at least one aberration term within each diopter..

118. A method according to claim 116, wherein the lens is provided by designing a lens that is capable of reducing at least one aberration term resulting from the wavefront sensing of the aphakic eye.

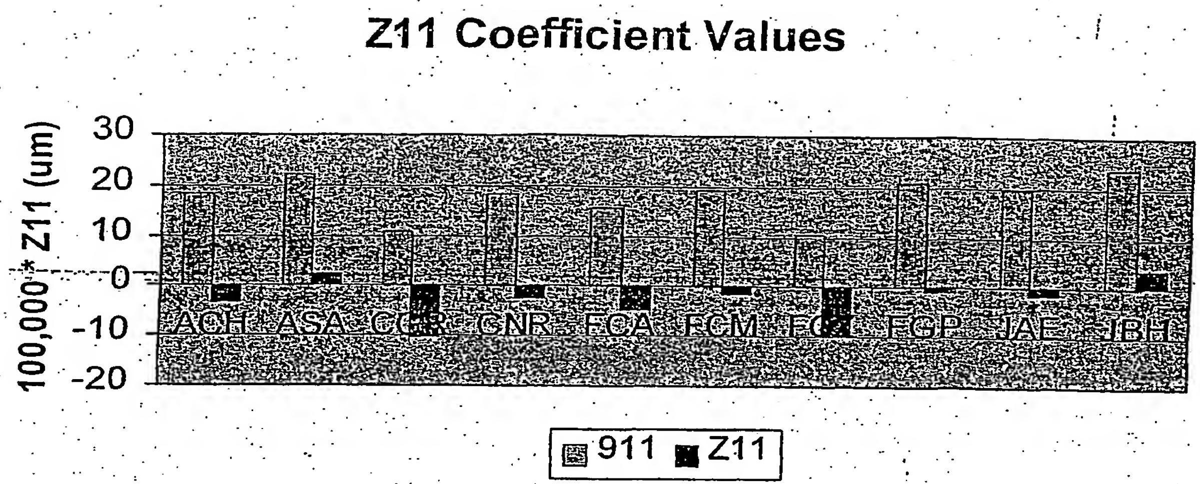

Z11 Coefficient Values
100,000 * Z11 (um)
30
20
10
0
-10
-20
ACH
ASA
CGR
CNR
FCA
FCM
FCZ
FGP
JAE
JBH
911
Z11

Fig. 1

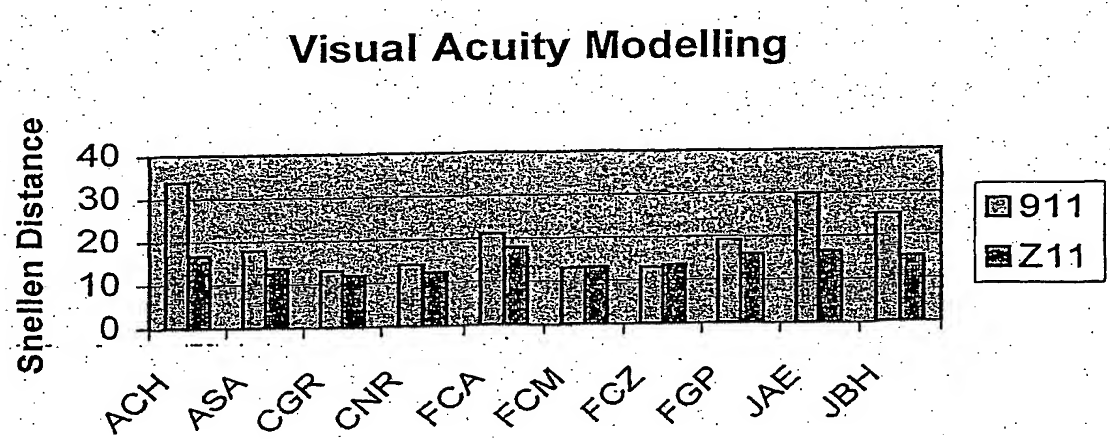

Visual Acuity Modelling
Snellen Distance
40
30
20
10
0
ACH
ASA
CGR
CNR
FCA
FCM
FCZ
FGP
JAE
JBH
911
Z11

Fig. 2

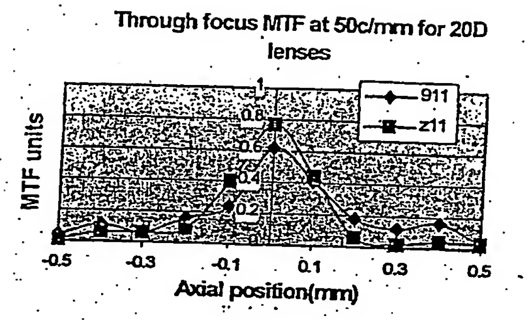


Fig. 3

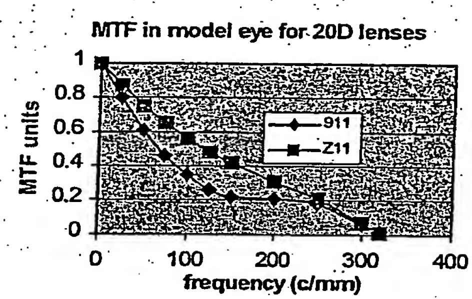


Fig. 4

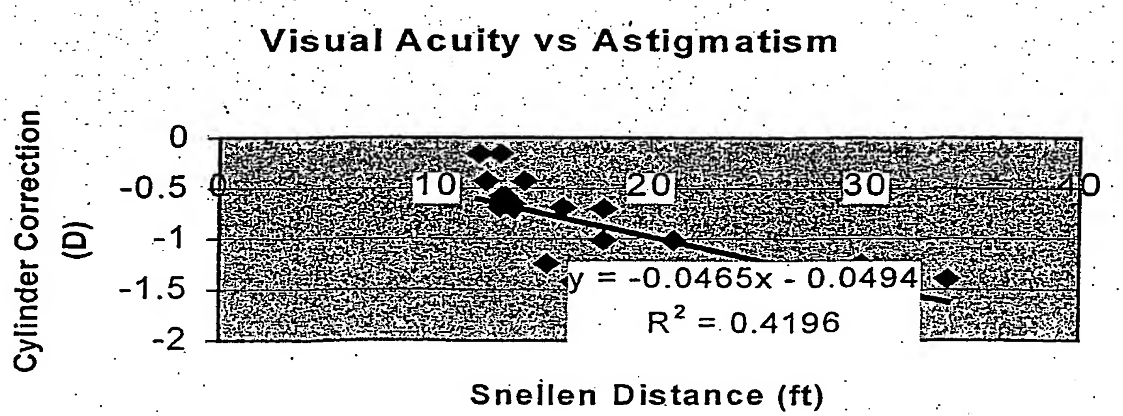


Fig. 5

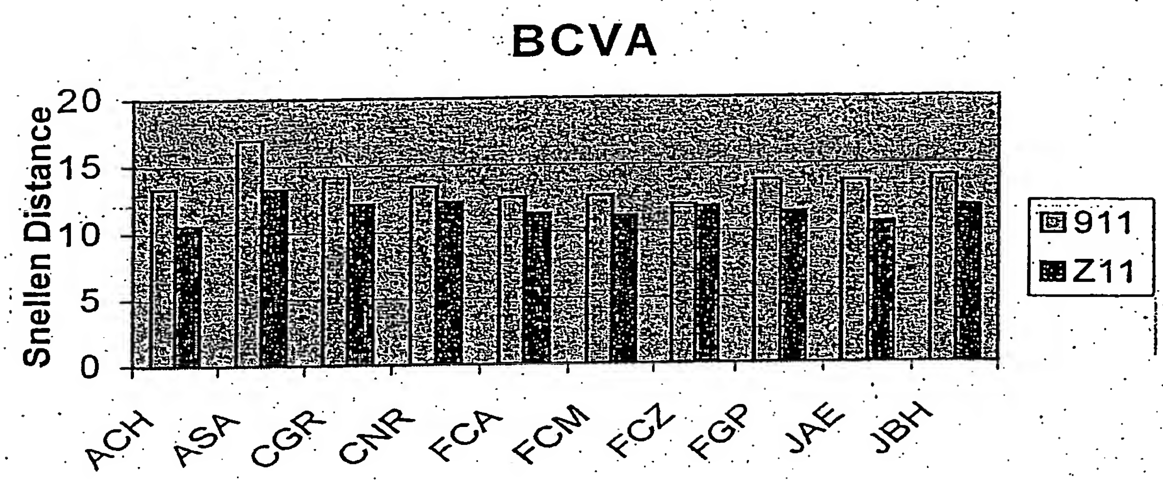

BCVA
Snellen Distance
20
15
10
5
0
911
Z11
ACH
ASA
CGR
CNR
FCA
FCM
FCZ
FGP
JAE
JBH

Fig. 6

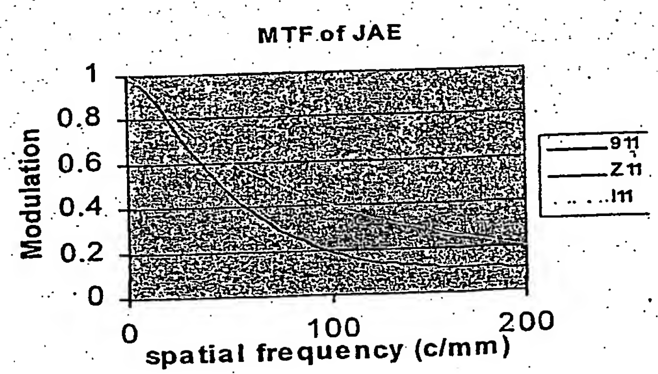

MTF of JAE
Modulation
1
0.8
0.6
0.4
0.2
0
0
100
200
spatial frequency (c/mm)
911
Z11
I11

Fig. 7

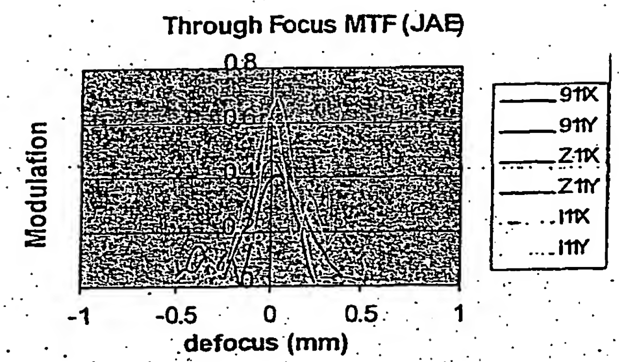

Through Focus MTF (JAE)
Modulation
defocus (mm)
-1
-0.5
0
0.5
1
911X
911Y
Z11X
Z11Y
I11X
I11Y

Fig. 8

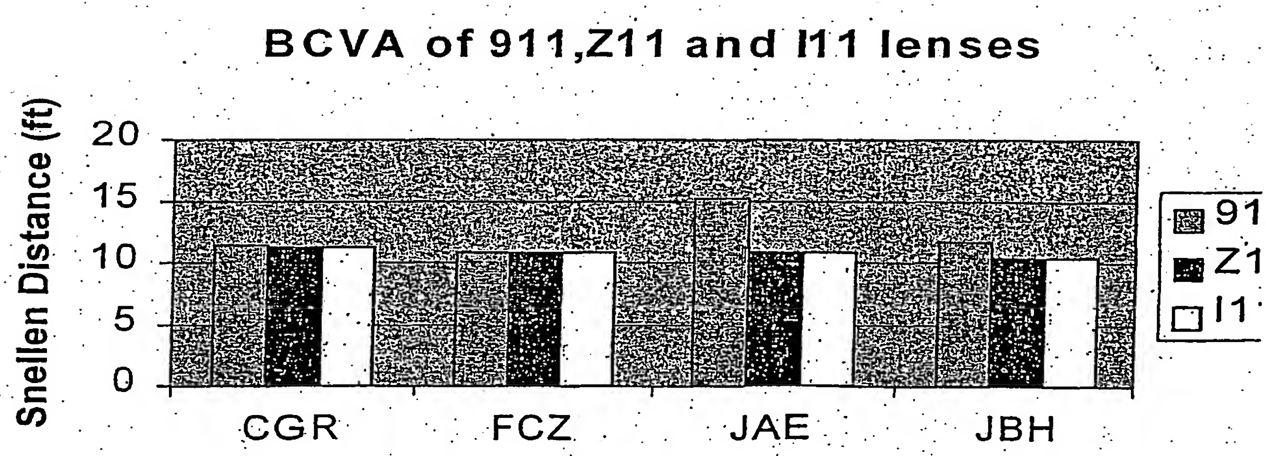

BCVA of 911,Z11 and I11 lenses
Snellen Distance (ft)
20
15
10
5
0
CGR
FCZ
JAE
JBH
91
Z1
I1

Fig. 9

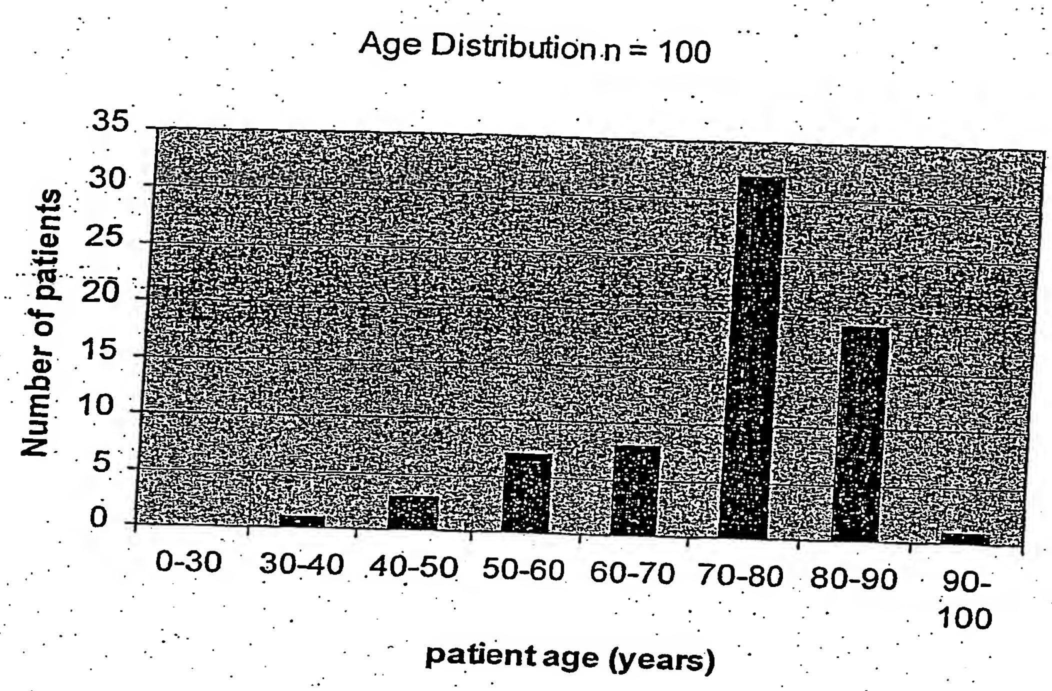

Age Distribution n = 100
Number of patients
35
30
25
20
15
10
5
0
0-30
30-40
40-50
50-60
60-70
70-80
80-90
90-100
patient age (years)

Fig. 10

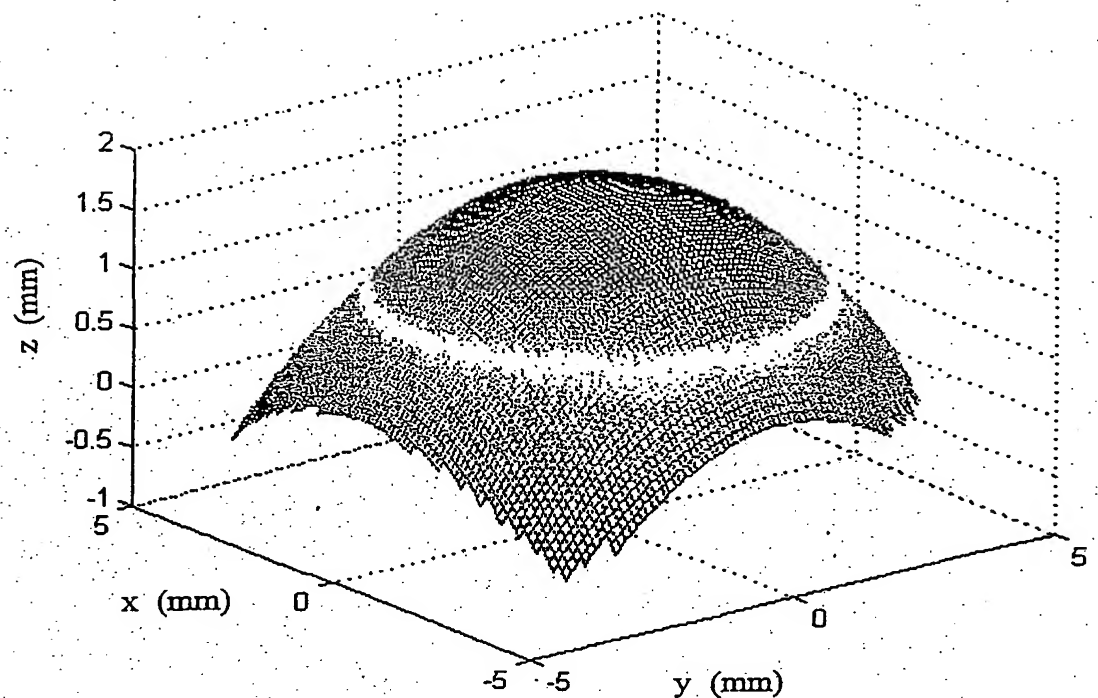


Fig. 11

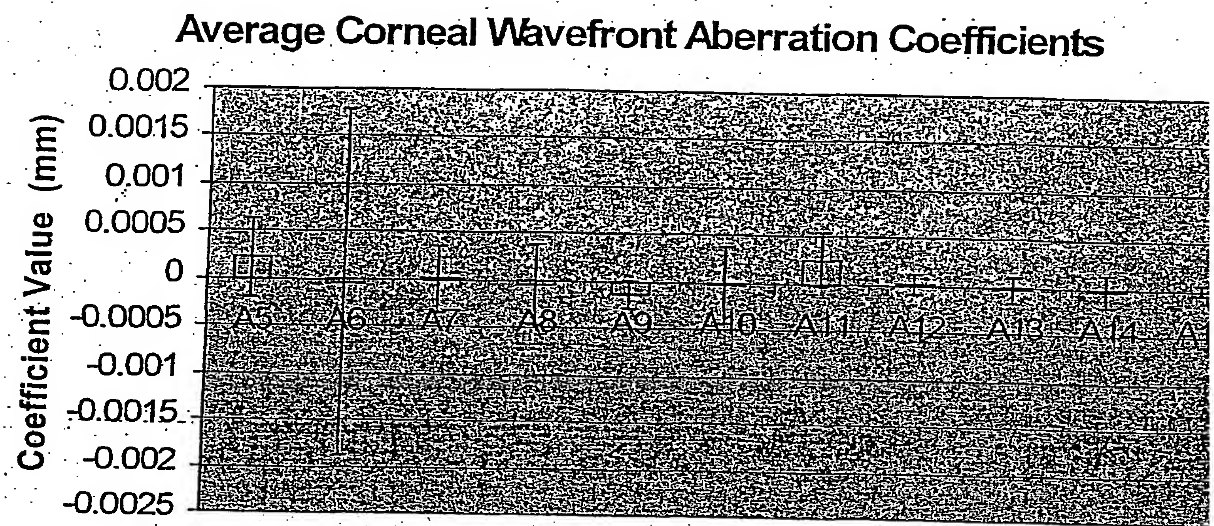

Average Corneal Wavefront Aberration Coefficients
Coefficient Value (mm)
0.002
0.0015
0.001
0.0005
0
-0.0005
-0.001
-0.0015
-0.002
-0.0025
A5
A6
A7
A8
A9
A10
A11
A12
A13
A14

Fig. 12

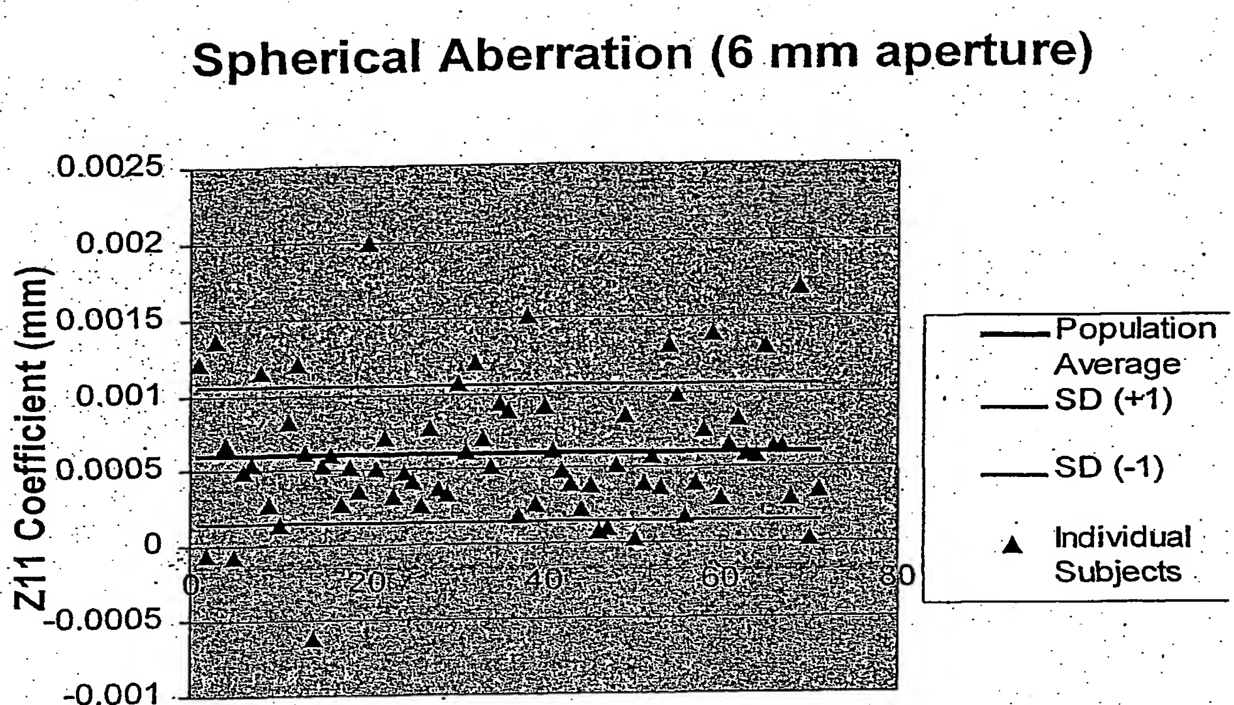

Spherical Aberration (6 mm aperture)
Z11 Coefficient (mm)
0.0025
0.002
0.0015
0.001
0.0005
0
-0.0005
-0.001
0
20
40
60
80
Population Average
SD (+1)
SD (-1)
Individual Subjects

Fig. 13

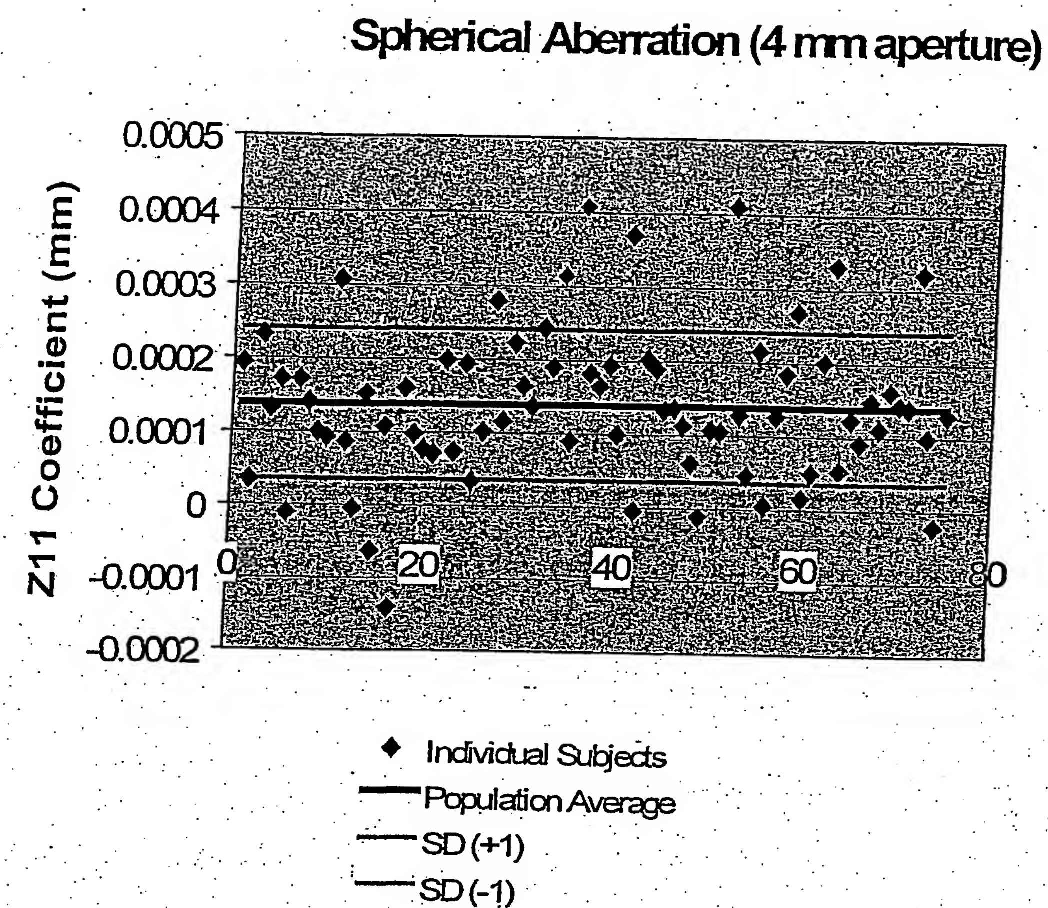

Spherical Aberration (4 mm aperture)
Z11 Coefficient (mm)
0.0005
0.0004
0.0003
0.0002
0.0001
0
-0.0001
-0.0002
0
20
40
60
80
Individual Subjects
Population Average
SD (+1)
SD (-1)

Fig. 14

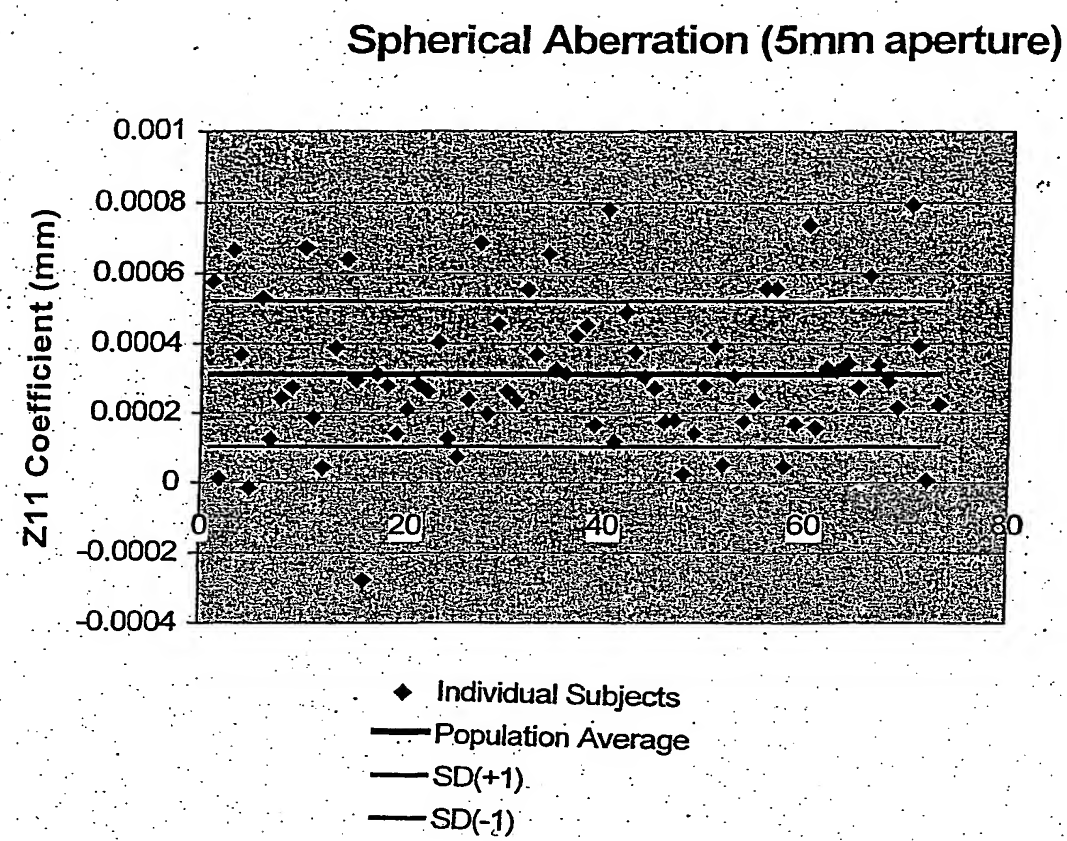
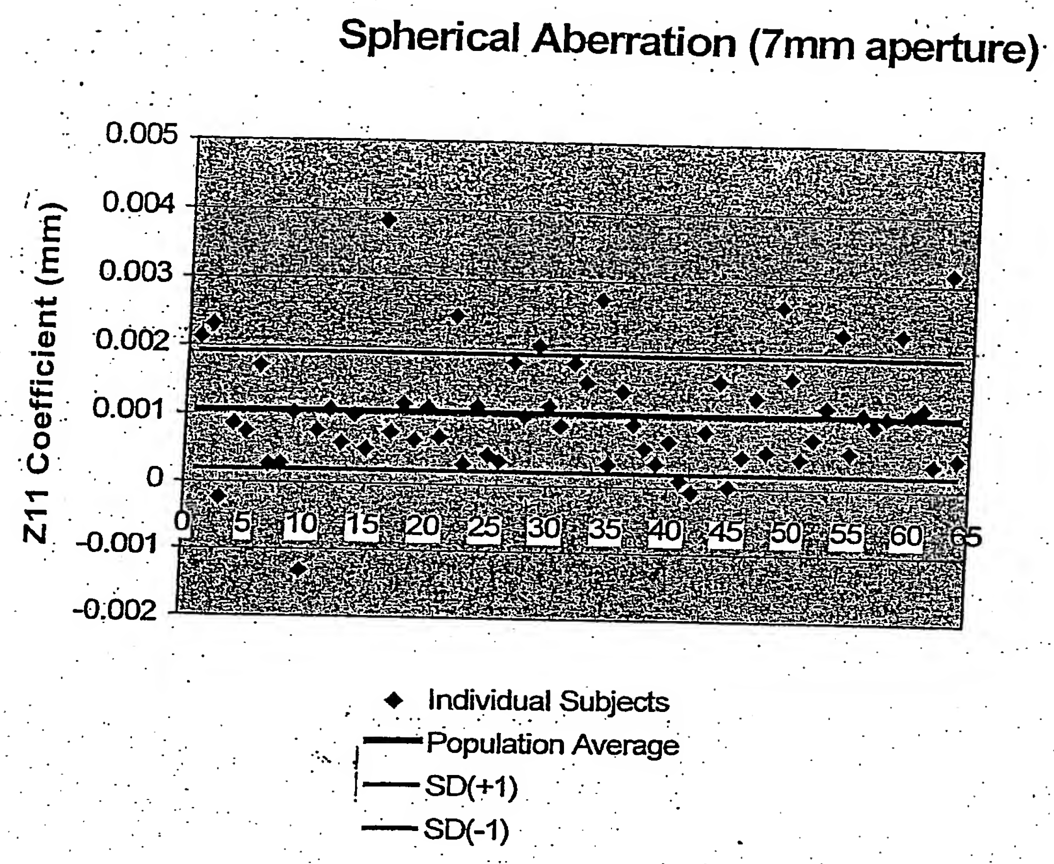

Spherical Aberration (5mm aperture)
Z11 Coefficient (mm)
0.001
0.0008
0.0006
0.0004
0.0002
0
-0.0002
-0.0004
0
20
40
60
80
Individual Subjects
Population Average
SD(+1)
SD(-1)

Fig. 15



Spherical Aberration (7mm aperture)
Z11 Coefficient (mm)
0.005
0.004
0.003
0.002
0.001
0
-0.001
-0.002
0
5
10
15
20
25
30
35
40
45
50
55
60
65
Individual Subjects
Population Average
SD(+1)
SD(-1)

Fig. 16

PCT/EP 01/06041

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7 A61F2/16 A61B3/107 G02C7/02

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
IPC 7 A61F A61B G02C

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, search terms used)

EPO-Internal, MEDLINE

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	DATABASE PUBMED 'Online! NCBI; SEITZ ET AL.: "Corneal topography" Database accession no. 10170448 XP002177903 abstract	1,4,7, 11-13, 17,25, 77,78, 82,105, 108-110
A		30,32, 34, 37-42, 55, 57-59, 61,62, 79,80, 83,86,88
	& CURR OPIN OPHTHALMOL, vol. 8, no. 4, August 1997 (1997-08), pages 8-24,	

-/--

[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

° Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

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"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
15 October 2001	31/10/2001

Name and mailing address of the ISA	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31-70) 340-2040, Tx. 31 651 epo nl, Fax: (+31-70) 340-3016	Neumann, E

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication,where appropriate, of the relevant passages	Relevant to claim No.
X	US 5 777 719 A (LIANG JUNZHONG ET AL) 7 July 1998 (1998-07-07) cited in the application	1,2,4,6, 8,11,12, 17-21, 25,26, 77-82, 105, 108-111
A	the whole document	30, 34-36, 38-40, 42-47, 49,50

Continuation of Box I.2

Claims Nos.: 68-70,71-76 (when depending on claim 68),77-91,93-104 (all in part and not depending on claim 92),92

Present claims 77to 104 relate to a product defined by reference to a desirable characteristic or property, namely an ophtalmic lens which is designed to obtain reduced aberrations.

The claims cover all products having this characteristic or property, whereas the application provides support within the meaning of Article 6 PCT and/or disclosure within the meaning of Article 5 PCT for only a ver: limited number of such products. In the present case, the claims so lack support, and the application so lacks disclosure, that a meaningful search over the whole of the claimed scope is impossible. Independent of the above reasoning, the claims also lack clarity (Article 6 PCT). An attempt is made to define the product by reference to a result to be achieved. Again, this lack of clarity in the present case is such as to render a meaningful search over the whole of the claimed scope impossible. Consequently, the search has been carried out for those part: of the claims which appear to be clear, supported and disclosed, namely those parts relating to the products, namely

expressing the cornea in mathematical terms so as to obtain aberration terms in order to produce an ophtalmic lens.

Present claims 68 and 92 relate to a product, namely an ophtalmic lens defined (inter alia)
in a formula
by reference to the following parameter(s):
: -1 < cc < Oonstant cc
P2: central lens radius R
P3: aspheric constants ad, ae
The use of these parameters in the present context is considered to lead to a lack of clarity within the meaning of Article 6 PCT. It is impossible to compare the parameters the applicant has chosen to employ with what is set out in the prior art. The lack of clarity is such as to render a meaningful complete search impossible. Consequently, the search has been restricted to all other searchable claims.

The applicant's attention is drawn to the fact that claims, or parts of claims, relating to inventions in respect of which no international search report has been established need not be the subject of an international preliminary examination (Rule 66.1(e) PCT). The applicant is advised that the EPO policy when acting as an International Preliminary Examining Authority is normally not to carry out a preliminary examination on matter which has not been searched. This is the case irrespective of whether or not the claims are amended following receipt of the search report or during any Chapter II procedure.

Patent document cited in search report		Publication date	Patent family member(s)			Publication date
US 5777719	A	07-07-1998	AU	723645	B2	31-08-2000
			AU	5806298	A	17-07-1998
			BR	9714178	A	29-02-2000
			CN	1245406	A	23-02-2000
			EP	0969760	A1	12-01-2000
			JP	2001507258	T	05-06-2001
			US	5949521	A	07-09-1999
			WO	9827863	A1	02-07-1998